

École polytechnique de Louvain

Design and prototyping of a robotic hip reeducation device

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Abstract

Nowadays, with more than 400 000 interventions per year in the United States, total hip arthroplasty (THA) has become one of the most successful surgeries of modern medicine [1]. In fact, it offers a last resort, when medications, neither the use of walking supports, improve the persistent pain and disability in everyday tasks.

In the context of the hip post-operative rehabilitation, Mr. Thanh Binh Nguyễn investigated the importance of the Gluteus Medius in the lateral stabilization of the pelvis. He noticed that insufficiency of this muscle usually manifested after a THA, impacts patient daily activities, such as walking. Thus, he believed that special attention should be given to the muscle during the rehabilitation program. For that purpose, the idea of building an automated solution, to replace the physically demanding and time-consuming conventional therapy, arose naturally out of his mind.

Even if several therapeutic devices are already commercialized, they are limited to passive movements. This is where this project came into play, to improve the physical recovery of hip abductors by adjusting the generated movements according to a patient's behavior. Hence, this thesis aims to provide a detailed design of an assistive device performing hip abduction and adduction exercises.

After a brief state of the art of rehabilitation practices, we noticed that the existing robotic solutions are difficult to transport and do not minimize the stresses inside the hip. For that purpose, this thesis includes the modeling of an exhaustive set of mechanisms, in the search of the most compact and less harming device to the weakened joint.

By reviewing the linear actuation technologies, selecting the required instrumentation tools and exploring the mechanical solutions minimizing bulkiness, we suggest a design solution.

The final prototype is a flattened transportable device which provides an accurate hip exercising shortly after the surgery. For its use, the device sits over the patient's bed (or therapy table) where the operated leg is mobilized by a support shell. Equipped with a belt-driven linear guide, the support moves back and forth, generating the abduction and adduction of the hip. In addition, the patient effort is collected with a miniature force sensor, in order to provide an adapted motor torque, through a real-time control.

Nevertheless, several aspects should be further investigated, mainly concerning the validation of the simulated models, as well as the design and selection of a suitable controlling strategy.

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Chapter 1

Introduction

1.1 Project history

In the context of his Master's thesis in 1992, Mr. Thanh Binh Nguyễn studied the importance of the *Gluteus Medius* in the lateral stabilization of the pelvis, especially after a hip surgery.

His research concludes that this muscle is essential for the static stabilization of the pelvis but equally important for dynamic activities such as walking. Thus, its insufficiency systematically results in a pathological gait. Moreover, the therapist was surprised that no automated solution assisting the abduction of the hip existed in the market. He believed that such a machine would bring clear value in the strengthening of the muscle in question since the conventional manual therapy is physically demanding and inaccurate.

Hence, he decided to build a prototype, named M-VITEC. The device would allow a passive and secure lower limb mobilization, generating isokinetic movements, all provided automatically with the help of a ball screw drive system. However, with an increasingly demanding job, he had to put aside his idea for a while.

More than two decades later, there was still no sign of such device commercialized, so he decided to bring the subject back to the table by contacting the UCLouvain in the search of engineers. As part of the course of "*Mechanical design in biomedical engineering*" (LMECA2855) in late 2017, my colleagues ¹ and I were asked to carry out this project. We met Mr. Thanh Binh Nguyễn, who shared with us his story and expectations.

After several months of development, we went through the preliminary stages of the project. Firstly, the technical requirements were reviewed and detailed in accordance with the therapist needs. Then, a special focus on the safety aspects and risks related to the use of M-VITEC were conducted through a risk analysis. In addition, we investigated the regulatory process of a medical device such as M-VITEC with the perspective of future commercialization. Finally, the main work was the design process where several technical solutions were explored by identifying their assets, but also their drawbacks, in order to obtain the most appropriate design for this specific clinical application. The followed methodology and final CAD prototype will be explained more deeply in section 5.1. After all, Mr. Thanh Binh Nguyễn was really pleased with the first advancements, motivating him to keep working in view of a physical model.

¹Alexandre Arnould, Julien Demey, Martin Guillaumie and Tanguy Nève

1.2 Thesis purpose & Composition

One of the principal reason responsible for muscular weakness around the hip joint is total hip arthroplasty (THA) (also known as total hip replacement). The *OECD* organisation reported more than 25 000 hip replacements in Belgium during the year of 2015 [2]. In addition, according to the Agency for Healthcare Research and Quality, more than 400 000 total hip replacements are performed each year in the United States, making THA is one of the most successful interventions in all of medicine [1]. As illustrated in Figure 1.1, partly due to the aging of the population, the number of hip replacement surgery continues to grow since 2000, in the countries belonging to *OECD*.

Moreover, a post-operative rehabilitation has proven to be inevitable for a healthier return to daily activities. However muscular deficits still manifest on a large amount of patient in the long term. Hence, this project addresses a global scale clinical need, by facilitating and refining the physical recovery of hip abductors.

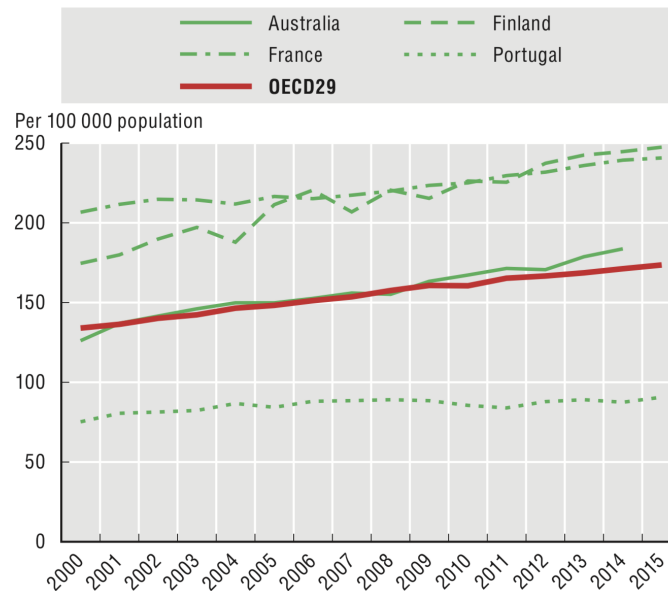


FIGURE 1.1: Hip replacement surgery trends, 2000 to 2015 (or nearest year) [2]

In the framework of this Master's thesis, we will start by giving an overview of the causes, the common surgical practices and the key steps of a conventional rehabilitation therapy regarding THA. Additionally, the role of robotics in the re-education of the lower limb is discussed, along with their limitations.

Then, the functions and characteristics of M-VITEC are presented via the technical requirements and a functional diagram. Afterward, we will go through the conceptual design which recaps the work accomplished previously and elaborates a more exhaustive study and classification of the possible actuation solutions, supported by a *Robotran* force analysis.

Further, we will focus on the "best" design solution by explaining the selection of the drive technology and outlining the instrumentation tools required for efficient and safe control of the machine. Finally, a detailed description of the CAD prototype is given, exposing the mechanical components and their assembly.

Chapter 2

Total hip arthroscopy

This first chapter briefly explains the morphology of the hip, as well as the problematic joint diseases affecting the daily lives of individuals and ultimately, forcing its replacement. For this purpose, the classic prosthetic hip is presented along with the main surgical techniques concerning its implantation. Finally, a first insight about the muscular impact caused by the main THA surgical approaches is given, based on a recent scientific research.

2.1 Hip anatomy & disorders

Along with the knee, the hip is one of the biggest joints of the human body and belongs to the family of ball-and-socket joints. Moreover, it enables the movements of the upper leg, while supporting the body weight in both static and dynamic situations.

As represented in figure 2.1, the socket is formed by the acetabulum, a concave surface located in the pelvis. Then, the head of the upper extremity of the femur, fits the acetabulum, forming the joint. Both bone surfaces are covered with a strong and lubricated layer, an articular cartilage, enabling them to slide smoothly. Several external ligaments and muscles en-capsule and provide stability to the joint, as well as generate the movements of the lower limb [3][4].

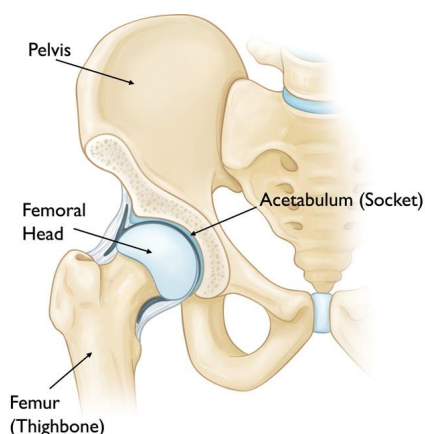


FIGURE 2.1: Healthy hip [3]

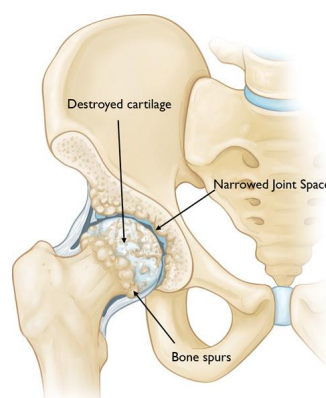


FIGURE 2.2: Hip affected by osteoarthritis [3]

The principal reasons to resort to a total hip replacement is chronic hip pain and joint stiffness, inducing motor disabilities. Some of the most common conditions provoking this kind of symptoms are osteoarthritis, rheumatoid arthritis or traumatic arthritis. Less frequently, hip fractures can also demand a complete hip replacement, happening most of the time in elderly people suffering from osteoporosis (brittle bones) [3] [5].

- **Osteoarthritis**, also called degenerative joint disease or “wear and tear” arthritis, occurs most frequently in the hands, hips, and knees in people 50 years of age and older. The cartilage cushioning the bones wears away, so the bones rub against each other, causing pain, stiffness, and swelling. In critical cases, it leads to reduced function and disability interfering with daily tasks and work [3] [6].
- **Rheumatoid arthritis** is an autoimmune disorder which occurs when your immune system erroneously damages your own body’s tissues. When the synovial membrane is affected, it inflates and thickens. Additionally, this chronic inflammation deteriorates the cartilage, leading stiffness and severe pain [7][3].
- **Post-traumatic arthritis** is caused by the wearing out of a joint due to a serious hip injury or fracture. Such an event can damage the bone and/or the cartilage, altering the mechanics of the joint and making it wear out faster. As the cartilage becomes damaged, it leads to hip pain and stiffness over time [3] [8].

These diseases can lead to pathological hip, as shown in Figure 2.2, where the cartilage wore out. The narrowed joint space made the bone surfaces to rub and become irregular, causing hip pain and joint inflexibility.

2.2 Hip prosthesis & THA surgical approaches

When the hip pain is acute, present while resting and everyday activities turn complicated, a safe and efficient solution is a total hip replacement. The procedure consists of removing the damaged bone and cartilage, replacing them with prosthetic components.

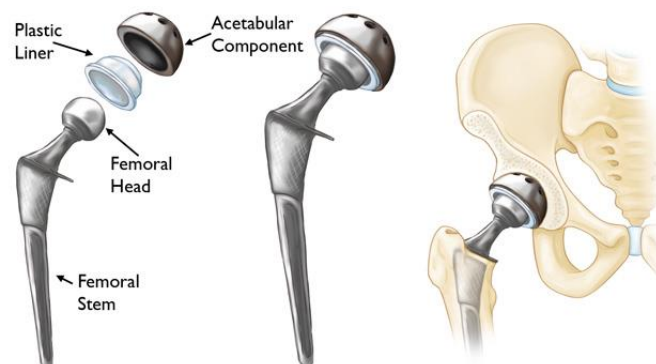


FIGURE 2.3: Prosthetic hip [3]

Figure 2.3 illustrate a classic hip prosthesis, which is composed of several parts. Firstly, a metal stem is inserted in the hollow center of the femur. Then, at its upper extremity, a ball replaces the damaged femoral head that was removed. Further, the wore cartilage surface of the socket is cleared and substituted with a metal socket. Screws or cement are usually used to hold the socket in place. Finally, a plastic, ceramic, or metal liner is placed between the femoral head and the socket, forming a smooth gliding surface [3] [9].

Around the globe, several surgical approaches are used to insert the hip prosthesis, but the following have been the most preferred by the medical community: the posterior approach, the direct lateral approach, and the anterior approach [10] (Figure 2.4).

- **A) Posterior approach (PA):** This approach is the most common surgical technique used worldwide. In fact, it provides adequate visualization of both the acetabulum and femur by coming from behind the patients' pelvis [11]. Thus, the surgeon has to cut several muscles and soft tissues to access correctly the hip, namely the *gluteus maximus* and the external rotators. These latter present an important role in postural stability and the abduction movement [12].
- **B) Direct lateral approach (DLA):** By attacking the hip laterally, this approach allows a very good exposure of the hip joint and is considered to have a low postoperative dislocation rate [13]. However, the *medius gluteus* and the *vastus lateralis* need to be split longitudinally during the intervention. Even though they are reconnected, the muscles are damaged and need precise therapy to regain their complete function. Especially, for the *medius gluteus*, which is the major abductor of the hip joint [14].
- **C) Anterior approach (AA):** This approach is the most recent and refers to surgeries accessing the hip from the front, usually through a smaller incision. This method is also called the muscle-splitting approach since the surgeon works between the muscles, with minimal or no muscle cutting by benefiting from the intermuscular intervals. Hence, this approach results on a faster rehabilitation by not having worry about the healing of the reattached muscles. In contrast, other complications are related to this method such as the high risk of injury of a large nerve just adjacent to the incision spot [15].

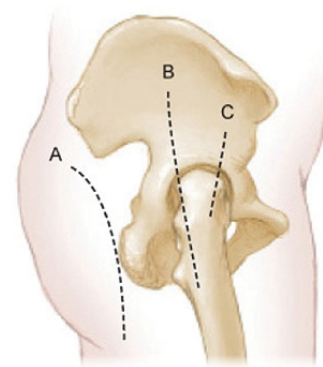


FIGURE 2.4: Surgical approaches: (A) Posterior, (B) Direct lateral and (C) Direct anterior.

In fact, all approaches have their unique advantages and disadvantages, but all can safely and successfully be performed for THA. The chosen approach depends on the patient condition, anatomy and the surgeon's experience. A wide range of complications can manifest during or after the intervention, such as hip dislocation, nerve injury, bleeding, intraoperative fracture, infection and abductor insufficiency. Each

technique presents different levels of invasiveness for the body, namely regarding tissue damage around the hip [16].

The Orthopaedic Research Centre of Trondheim University Hospital in Norway led a study that reflects the impact of the mentioned surgical methods on muscles, by comparing muscular strength after a total hip replacement [14]. Through this research, 60 patients, subjects of different approaches of THA, were submitted to several tests, including abduction strength tests.

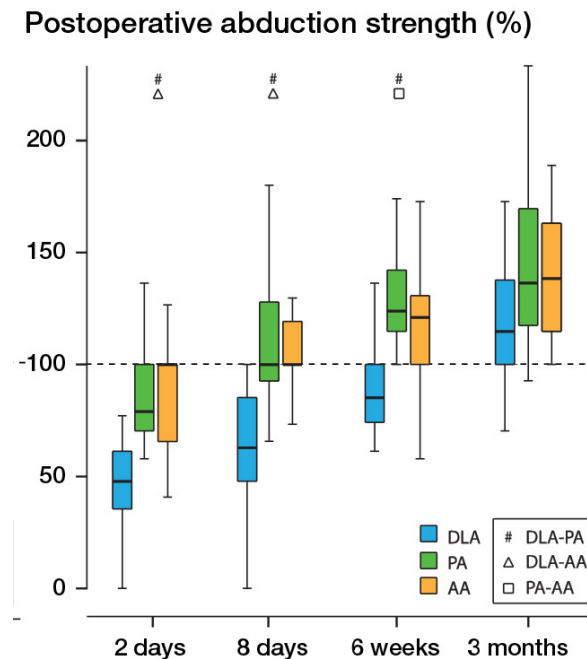


FIGURE 2.5: Postoperative abduction strength compared with preoperative values (100%) after the direct lateral approach (DLA), the posterior approach (PA) and the anterior approach (AA) [16]

Figure 2.5 reports the abduction strength tests results for a follow-up of 3 months after performing the three surgical techniques. As we can see, 2 days after the intervention, all groups of patients present a loss of post-operative abduction strength. In particular, shortly after the operation, patients that underwent a DLA approach ($< 50\%$ for 2 days and around 60% for 8 days), present statistically significant less abduction strength compared to the other two groups. In other words, it confirms what was previously described, DLA is more invasive for the abductor muscles. Throughout the evaluation period, PA causes less muscular weakness, but after three months there is no significant difference in the abduction strength for the three approaches. Therefore, each technique damages differently the muscles around the hip, but they all reach the muscular strength at the end of the testing period. Nevertheless, this study demonstrates that the operated legs were weaker than the untouched legs for the three groups of patients.

Hence, this constitutes an indicator that THA has post-surgical consequences, especially concerning muscular performance. As will be discussed in the next chapter, the final clinical outcome of a THA heavily depends on the ingredients of the post-operative rehabilitation program.

Chapter 3

Hip rehabilitation

The rehabilitation protocols following total hip arthroscopy are recognized as an integral component of the final clinical outcome of the surgery.

This chapter highlights the main goals and current practices for the postoperative recovery, along with their limitations. These are usually related to the increase in quantity, complexity, and diversity of surgical procedures performed in an evolving patient population. Finally, across an overview of existing lower limbs rehabilitation devices, we will investigate their potential and technical drawbacks in the context of hip postoperative rehabilitation therapy.

3.1 Goals & practises of hip rehabilitation

Despite the variations and inconsistencies in the existing therapies after THA, there are principles commonly applied in the majority of rehabilitation protocols.

The main goals of the rehabilitation therapies are to regain the natural range of motion (ROM), muscle strength, endurance, and functional capacity by avoiding postoperative complications [17]. To ensure safety and effectiveness, the therapy should be performed under the supervision of a medical expert. He recommends the most adequate program, according to its clinical experience, the type of surgical procedure and patient's clinical records, by targeting the weakened muscles.

Mainly, the muscle groups that are affected and so, concerned in rehabilitation programs, are the abductors, adductors and gluteal muscles [18]. In addition, the rehabilitation must be continuously adapted with respect to the recovery progress of the hip joint.

Usually, the rehabilitation process starts before the surgery with a preoperative consulting session, which has proven to be beneficial for the patient. It is the opportunity to explain the demands and time investment involved in the postoperative program. It allows the clinician to identify the patients less compliant with the therapeutic protocol and intervene in order to stimulate patient engagement. Thus, the patient becomes less apprehensive allowing a faster and more efficient start of the treatment directly after the surgery [17].

Shortly after the operation, the physical therapy begins in order to minimize tissue inflammation, control pain and provide soft activities, optimizing the healing process, stimulating adequate blood circulation and enabling joint ROM [17]. In practice, this phase is **passive**, performed manually by the therapist who mobilizes and

moves repetitively the leg, without effort from the patient [19]. The damaged tissues around the hip start to restore, such as the joint capsules, ligaments, muscles, and the skin. Their proper function is essential to maintain a stable and an upright posture while doing daily activities. At an early stage after the procedure, the proprioceptive function of the tissues is disrupted, which leads to a balance impairment, increasing fall risk, especially in elderly people [20]. Passive activities should start within the 24 hours after the surgery and are typically prescribed for at least one to two weeks [17].

Very soon, as the patient progresses, the exercises become muscularly demanding and require higher autonomy. The patient will start a phase composed of **active** exercises improving flexibility and muscular strength [19]. For instance, the practice of stationary bike is often used for repetitive ROM, as well as devices performing a continuous passive range of motion exercises. These activities should be performed several times a day. Additional exercises commonly prescribed within the first week after the surgery include ankle ROM, quadriceps and gluteal isometric exercises, as well as pelvic exercises [17]. They should be maintained for at least four weeks and practice on a daily basis.

It is primordial to insist on the gluteal muscles strengthening, especially the *gluteus medius* and hip abductors due to their role of frontal plane stabilizer of the pelvis in functional activities [21]. Therefore, for a quick and complete recovery is fundamental to include exercises requesting this group of muscles.

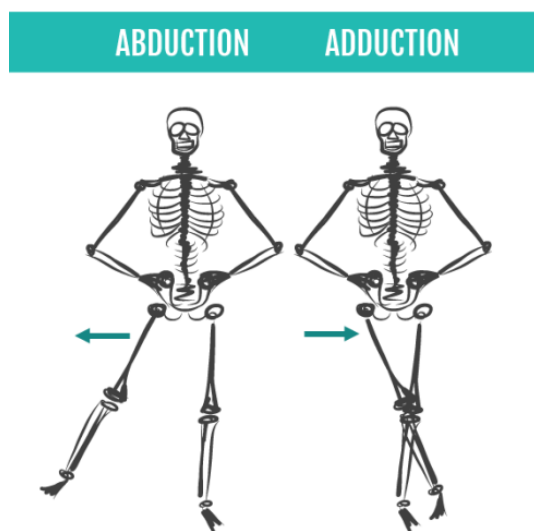


FIGURE 3.1: Hip abduction and adduction movements [22]

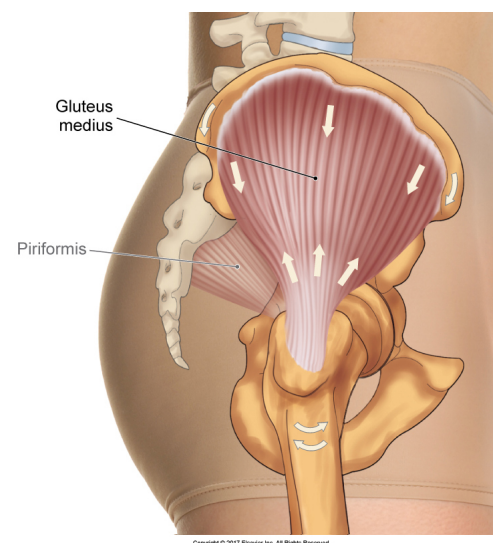


FIGURE 3.2: Lateral view of the right *gluteus medius* [23]

A well-known and efficient exercise to reeducate the mentioned muscles is the **abduction** and **adduction** of the lower limb. The abduction is any motion of the limbs or other body parts that pulls away from the sagittal plane of the body. Adduction is the opposite movement which brings a body part towards the sagittal plane [24]. For instance, the motion moving away the leg to the side is called hip abduction, and the opposite is hip adduction (Figure 3.1).

As illustrated in Figure 3.2, hip abduction is actuated mainly by the *gluteus medius*, which pulls the upper extremity of femur generating the rotation of the leg around the hip joint. During adduction, the *gluteus medius* relaxes and adductors muscles contract leading to the reverse movement [25].

The last phase of rehabilitation starts when the normal levels of strength, ROM and flexibility are reached, without causing any pain, usually not before 6 weeks after the surgery [26]. At this stage, the patient performed **resistive** and weight-bearing exercises, with a special focus on the improvement of proprioception and coordination to achieve a regular gait [19]. These exercises are particularly important for top athletes seeking to perfectly harmonize their movements, allowing them to regain their best form.

For an easy-going transition from partial to full weight-bearing activity, aquatic mobile exercises are a commonly used practice [17].

Figure 3.3 gives an estimation of the duration of each phase of the physical therapy after a THA. However, note that the recovery time after a THA differs vastly from patient to patient, mostly depending on how regularly and seriously they perform the physical exercises. Some may take 6 months to recover, while others may recover in less than 5 weeks [27].

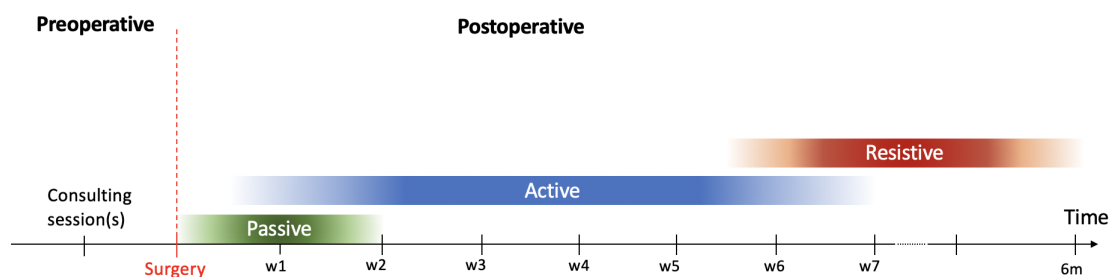


FIGURE 3.3: Physical rehabilitation time-line estimation

3.2 A limited clinical outcome

Most of the patients report better overall quality of life after THA, nevertheless, in a long term perspective, globally functional deficits still persist [28]. With THA becoming more and more popular on the global scale and the constant emerging of new surgical approaches, there is a need to emphasize patient-specific rehabilitation strategies improving long-term physical function. Even if THA is mainly performed in elderly people, the operation of a younger and healthier population is becoming more and more common, demanding a faster comeback to their active life [17].

Moreover, several studies report a lack of high-quality evidence for THA rehabilitation. The efficiency of treatment protocols and their clinical outcomes are hard to estimate. The current guidelines are mostly based on basic science and clinical experience, instead of individual characteristics and the applied surgical procedure [29][30].

Many researches have noticed permanent functional impairment after THA, namely, leading to a pathological gait. As already claimed, it is known that the hip muscle capacity is deteriorated early after THA, independently of the used surgical technique [31]. In particular, abductors strength is severely affected which limits postoperative activities of daily living [32]. Several studies have tried to evaluate the impact of static and dynamic abductor impairment in long-term physical functionality, in people with a prosthetic hip [20][33]. In reality, both abductor strength and gait ability were significantly associated with physical function more than 1 year after the surgery. An early dysfunction of abductors will lead to permanent balance and gait deficiencies affecting not only the hip but also the back and the knee.

A common consequence of weakened abductor muscles is the manifestation of an asymmetrical gait, known as the Trendelenburg gait. As shown in Figure 3.4 (A), during the stance phase (60% of the gait cycle), the *gluteus medius* contracts to maintain both hips at the same level. When the right *gluteus medius* is ineffective, the left pelvis side drops during the right stance phase (B), causing a characteristic limping [34].

This complication may suggest that rehabilitation may be more effective in the early stages after surgery, by emphasizing abductors reinforcement and coordination.

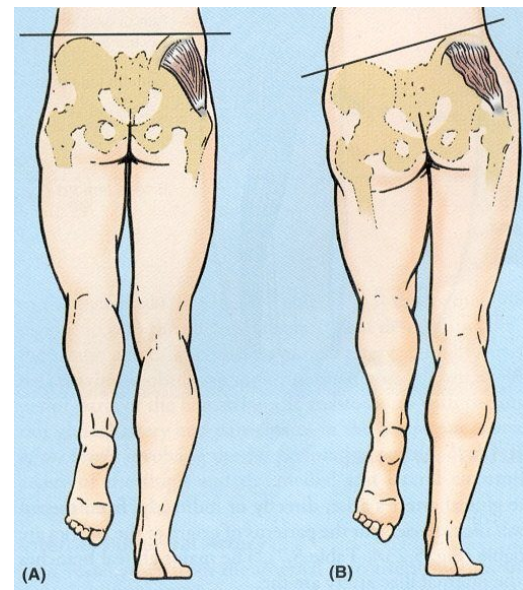


FIGURE 3.4: (A) Normal gait vs. (B) Trendelenburg gait [35]

3.3 Therapeutic robots for hip rehabilitation

The hip rehabilitation therapy is composed of repetitive physical movements which require manual mobilization, time and patience from the physiotherapist. Therefore, this long process turns time-consuming and physically demanding for physiotherapists. In addition, rehabilitation centers usually experience a lack of availability to attend various THA patients simultaneously [19]. Thus, limited time and attention can be given to each patient.

These are some of the reasons that spark the interest of the medical community to develop therapeutic robots in the field of rehabilitation.

In fact, the number of studies about robots for rehabilitation purposes has significantly increased, especially in the last decades [36].

Robots for lower limb rehabilitation can be divided into two main groups: gait rehabilitation devices and stationary systems.

3.3.1 Gait rehabilitation devices

This first type of devices helps the patient to recover the loss of function due to neurological disorders, accidents or after injury affecting their walking abilities. They substitute the physically challenging task of guiding the patient's legs during a conventional locomotion therapy. The most popular systems are powered exoskeletons in a fixed frame with a treadmill (Figure 3.5), that can actuate body joints separately all along the gait cycle. Several treadmill gait trainers are now available in the market such as, *Lokomat* [37] and *Walkbot* [38]. These robots can support the total body weight and the movement of patients.

Thus, they are beneficial for severely impaired patients who experience muscular paralysis, for instance, to treat post-stroke effects.

However, THA patients present lighter muscular deficit and only around the hip, thus the motor assistant provided by such robotic gait trainer might be inadequate. Moreover, the high cost of such devices is a serious limitation, particularly in developing countries, where only a few individuals have profited from these machines [40].

Lighter and easily wearable hip assistive devices like *Honda Walking Assistive* (HWA), presented in Figure 3.6, could be an effective solution to correct gait asymmetry. A recent study demonstrates their positive impact on the rehabilitation of early gait training in a patient with a prosthetic hip. In fact, it contributes to the improvement in hip function and gait pattern [41].



FIGURE 3.5: Lokomat by Hokoma [39]



FIGURE 3.6: Honda Walking Assistive Device [42]

Gait rehabilitation devices could be useful for THA patients whose hip ROM and muscular strength are sufficiently restored to walk freely but still experience a gait anomaly. Hence, these systems are effective in an advanced phase of physical therapy.

3.3.2 Stationary rehabilitation devices

Stationary systems are designed to exercise the human lower limb without walking. By mobilizing the limb, the machine performs repetitive and guided movements aiming to improve progressively joint mobility and muscular strength, by reducing the physical workload of rehabilitation [19].

The first version appeared in the late 70s as "Continuous Passive Motion"(CPM) devices and is intended to act in the initial stages of rehabilitation. Their impact was promising, considering that they seem to be painless, stimulate the regeneration of articular tissues and prevent joint stiffness [43] [44]. Hence, CPM devices are widely used in medical centers and hospitals, replacing the physiotherapist for simple passive exercises.

The more advanced model by *Kinetec*, provide hip and knee flexion/extension, but also the combination of hip abduction with flexion/extension in a lying down position (Figure 3.7).

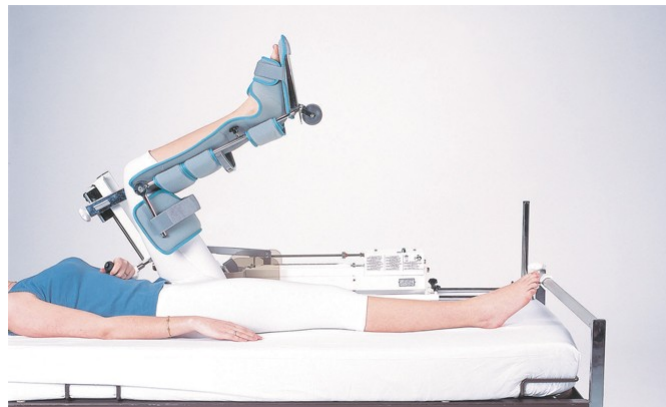


FIGURE 3.7: Side view of Kinetec 9081 Hip CPM device [45]

However, such systems are not responsive to the patient's behavior, since they are not equipped with sensors. In fact, when performing rehabilitation exercises, some patient move involuntarily their extremities, due to reflexes. Thus, the device and the patient motions are sometimes not coordinated, leading to undesirable physical strains. For safety reasons these systems are not the most suitable for physical therapy [46] [19].

Stationary rehabilitation devices must be more intelligent by modulating the movement trajectories and forces according to patient behavior. There are several prototypes recently developed, trying to address this issue, by means of real-time measurements of limb position and forces.

For instance, HipBot and Physiotherabot (Figure 3.8) prototypes performed combined hip abduction/adduction and flexion/extension. As a CPM device, they generate passive movements, but also active and resistive exercises, fundamental to strengthen the hip joint after a THA. Experiments ran on healthy subjects have demonstrated that these robots can cover the required ROM and correctly imitate the manual exercises performed by the physiotherapist [19] [47].

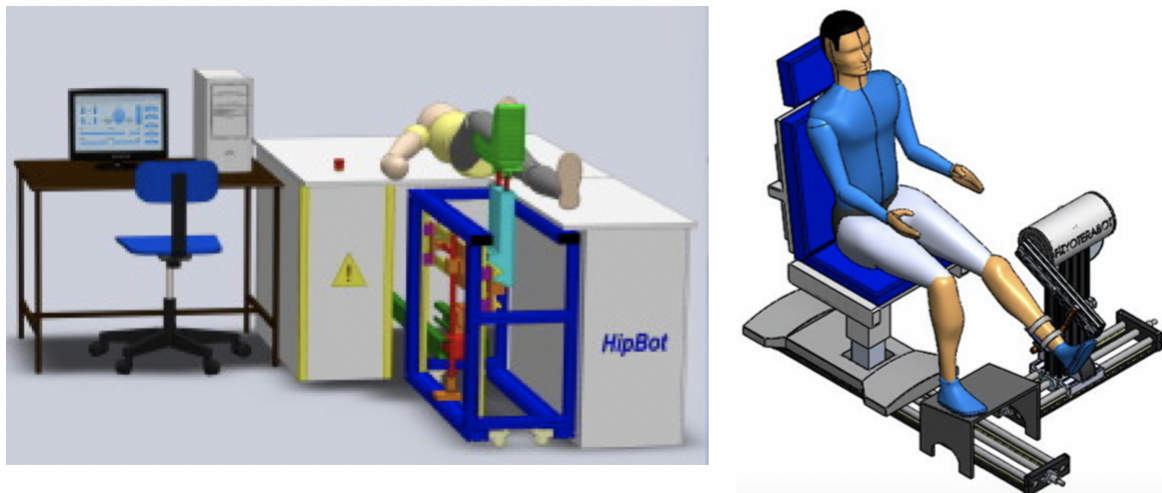


FIGURE 3.8: CAD models of Hipbot (left) and Physiotherabot (right) [19] [48]

Unfortunately, at each use, the patient must be positioned in the device so that his hip joint is precisely aligned with the device actuated joint. This results in a considerably long positioning phase, which is carefully handled by the physiotherapist according to the patient's leg morphology [19]. In fact, a misalignment between the prosthetic joint and the device axis of rotation generates hip strains that can surely harm the patient and lead to an uncomfortable man-machine interaction [49].

Furthermore, these prototypes are not mobile machines, so the patient needs to move to the room where they are permanently installed, being not practical, especially right after the surgery.

Finally, only Hipbot is capable to perform exercises for both legs, but the setup appears to be complicated for a physiotherapist, in a conventional medical environment.

These current shortcomings stimulated the interest of developing a new stationary rehabilitation device, presented in the following chapter.

Chapter 4

M-VITEC characterization

This chapter introduces M-VITEC, a stationary rehabilitation device assisting hip abduction while attempting to overcome the constraints of conventional therapies. First, are reported the potential benefits of this project for the clinical community. Additionally, the detailed set of technical requirements, followed by the functional diagram of the machine, are described.

4.1 Project potential

After investigating what had already been done for lower limb rehabilitation, the current surgical practices, and the expected clinical outcomes, we were able to estimate the potential of this project. Thus, here is a non-exhaustive list of the aimed benefits:

- Generate **abduction and adduction** hip movements thanks to a simple and safe system.
- **Workload relief** by automatizing physically demanding and repetitive exercises necessary at different phases of the rehabilitation process.
- Possibility to treat **multiple patients** simultaneously, thus improve problems related to the lack of medical staff and patient satisfaction.
- Provide a more **informed therapy** thanks to a continuous recording and saving of patient performance (ROM, developed force, etc.) enabling a closer follow-up.
- Perform **precise and controlled movements** (isokinetic exercises) practicing effectively the hip muscles, thus leading to a better clinical outcome.
- Unlike current therapeutic devices, the hip fully guides the movements which **minimize efforts** within the prosthetic joint.
- Can be **easily transported** to the patient location (hospital room, physiotherapy center, home, etc.), allowing a fast start of the rehabilitation program and a decrease in relative costs.

Naturally, these aspects could potentially be achieved if the device works as expected and the design requirements are fulfilled. We have to keep in mind that the use of such a device can put the patient in a variety of dangerous situations. A basic risk analysis related to the previous version of M-VITEC, done for the LMECA2855

course, can be found in the Appendix B (still valid for the prototype developed in this thesis). It provides some of the possible risks arising while an individual interacts with the device. This type of analysis is fundamental for the approval and commercialization as a medical device [50]. Thus, it is important to keep it in mind throughout all the design process.

4.2 Technical requirements

The design of a new device always implies a precise and exhaustive set of technical requirements established in the early stages. These describe the functions and features that need to be satisfied by the final product.

In reality, the technical requirements established in the frame of the LMECA2855 course (November 2017) have almost remained the same since the medical needs did not evolve in such a short period of time.

Nevertheless, the preliminary investigation presented in this thesis, allows us to understand the potential of such a device for the physiotherapist routine and its positive impact in THA clinical outcome.

In addition, we grasped the technical aspects to focus on all along the design process in order to bring a solution that could stand out from the existing therapeutic devices. Thus, the experienced physiotherapist insights and the state of the art regarding THA, provided us a good baseline to validate the previously set requirements.

The technical requirement that the device ultimately needs to comply with are presented in Table 4.1.

Firstly, the main functions are reported along with their detailed sub-requirements. Thereafter are listed the constraints and constraint requirements which are not directly related to any of the main functions. The source of each item is either the physiotherapist, Thanh Binh Nguyen (TBN) or the engineering team (ET) composed of the students involved prior in this project.

Technical Requirements - Version 3 - March 2019		
Date	Source	Main functions (MF)
02/10/17	TBN	MF 1 - The device assists hip abduction and adduction through three distinct operating modes.
05/03/19	TBN,ET	MF 2 - The device records and stores forces and amplitudes, in real time.
02/10/17	TBN	MF 3 - The device is placed between the consultation table and the patient's lower limb.
Date	Source	Function requirements (FR)
05/03/19	TBN,ET	FR 1.1 - The device has three distinct operating modes: • Passive mode : The patient is passive, the device applies the necessary force for the whole movement to reach a desired trajectory. • Active mode : The patient attempts the movement, the device supplies an assistive force to reach a desired trajectory. • Resistive mode : The device applies a force opposed to the patient's movement to reach a desired trajectory.
02/10/17	TBN	FR 1.2 - The movement is performed at constant angular speed for each of the 3 modes between $[0^\circ/\text{s} - 60^\circ/\text{s}]$ by steps of $10^\circ/\text{s}$ with an accuracy of 2° .
02/10/17	TBN,ET	FR 1.3 - The movement is composed of a succession of hip abductions and adductions. The other degrees of freedom are blocked.
09/10/17	ET	FR 1.4 - The movement amplitude is adjustable, between $[0^\circ - 60^\circ]$ by the physiotherapist, before the use of the device.
09/10/17	ET	FR 1.5 - The maximal angular acceleration/deceleration are $90^\circ/\text{s}^2$ and $-90^\circ/\text{s}^2$ respectively.
05/03/19	ET	FR 1.6 - The maximal torque generated on the hip is 50 N.m.
05/03/19	TBN,ET	FR 2.1 - Two types of measurement are made: • Movement angular amplitude (from the sagittal plane to the current position). • Force generated by the patient on the device.
02/10/17	TBN,ET	FR 2.2 - Measurements are taken in all three modes.
02/10/17	TBN	FR 3.1 - The patient's posterior rests on the examination table or bed.
02/10/17	TBN	FR 3.2 - The device holds the patient's feet in a vertical position (aligned with sagittal plane) avoiding any internal or external rotation of the hip.
09/10/17	TBN, ET	FR 3.3 - The maximal angle between patient's lower limb and the table is 15° (sagittal plane) avoiding any significant flexion of the hip.
09/10/17	TBN, ET	FR 3.4 - The device carries the lower limb, but guidance is ensured by the hip.
Date	Source	Constraints (C)
02/10/17	TBN	C 1 - The device is ergonomic.
02/10/17	TBN	C 2 - The device ensures patient's safety.
02/10/17	TBN	C 3 - The device is versatile.

02/10/17	TBN	C 4 - The device is portable.
02/10/17	TBN	C 5 - The device works with electric energy.
09/10/17	TBN, ET	C 6 : The device is competitive in the market.
Date	Source	Constraint requirements (CR)
02/10/17	ET	CR 1.1 - Patient height: between 130 and 200 cm.
09/10/17	ET	CR 1.2 - Maximal patient weight: 100 kg.
09/10/17	ET	CR 1.3 - Device maximal weight: 30 kg
09/10/17	TBN, ET	CR 2.1 - The device provides feedback to the physiotherapist to avoid hurting the patient. An alert system is activated when abnormal forces are detected.
02/10/17	TBN, ET	CR 2.2 - The device is fitted with an emergency stop button accessible for both the physiotherapist and the patient. The button will immediately stop any force acting on the patient's limb.
02/10/17	TBN	CR 2.3 - The device automatically stops when forces are too important. The maximal force is implemented by the physiotherapist before the device is used.
02/10/17	TBN	CR 3.1 - The device can be installed on any flat examination table or clinical bed at least 60 cm large and 160 cm long, with a maximal height of 120 cm.
09/10/17	TBN,ET	CR 3.2 - Symmetrical use: The device can be used for both the right and left lower limbs.
02/10/17	TBN	CR 3.3 - The healthy lower limb remains free. Only the injured lower limb is fastened to the device.
02/10/17	TBN	CR 4.1 - The device is movable by hand or with a special cart similar to what can be found in hospitals or clinical centers to move equipment around.
09/10/17	ET	CR 4.2 - The device can be installed and manipulated by a single person (i.e. the physiotherapist).
02/10/17	TBN	CR 5.1 - The device is pluggable into a wall outlet, functioning with 220-240V, 50 HZ, Standard EU electrical power network.
09/10/17	TBN, ET	CR 6.1 - The manufacturing price per unit must not exceed, if possible, 10.000 €.
05/03/19	ET	CR 6.2 - The device complies with the European regulations (Directive 93/42-CEE) for medical devices.

TABLE 4.1: Technical requirements - Version 3 - March 2019

4.3 Functional Diagram

After setting the technical requirements, a useful tool to visualize the flow of the operations is a functional graph. The functional diagram shown in Figure 4.1, is a block diagram gathering all the actions taken by the device, but also the interactions with the user during the use of the system.

Each block corresponds to one operation, either performed by an individual (health practitioner, patient, etc.) (dotted block) or the device itself (continuous line block). Black arrows stand for the natural sequence of actions during the use of the machine, while red arrows indicate all the paths leading to an emergency stop. Moreover, the utilization or recording of data is represented by the green rounded blocks. Finally, next to operation blocks are smaller dotted blocks referring to their related technical requirement from Table 4.1.

Note that this functional diagram was strongly inspired by the version established during LMECA2855 course.

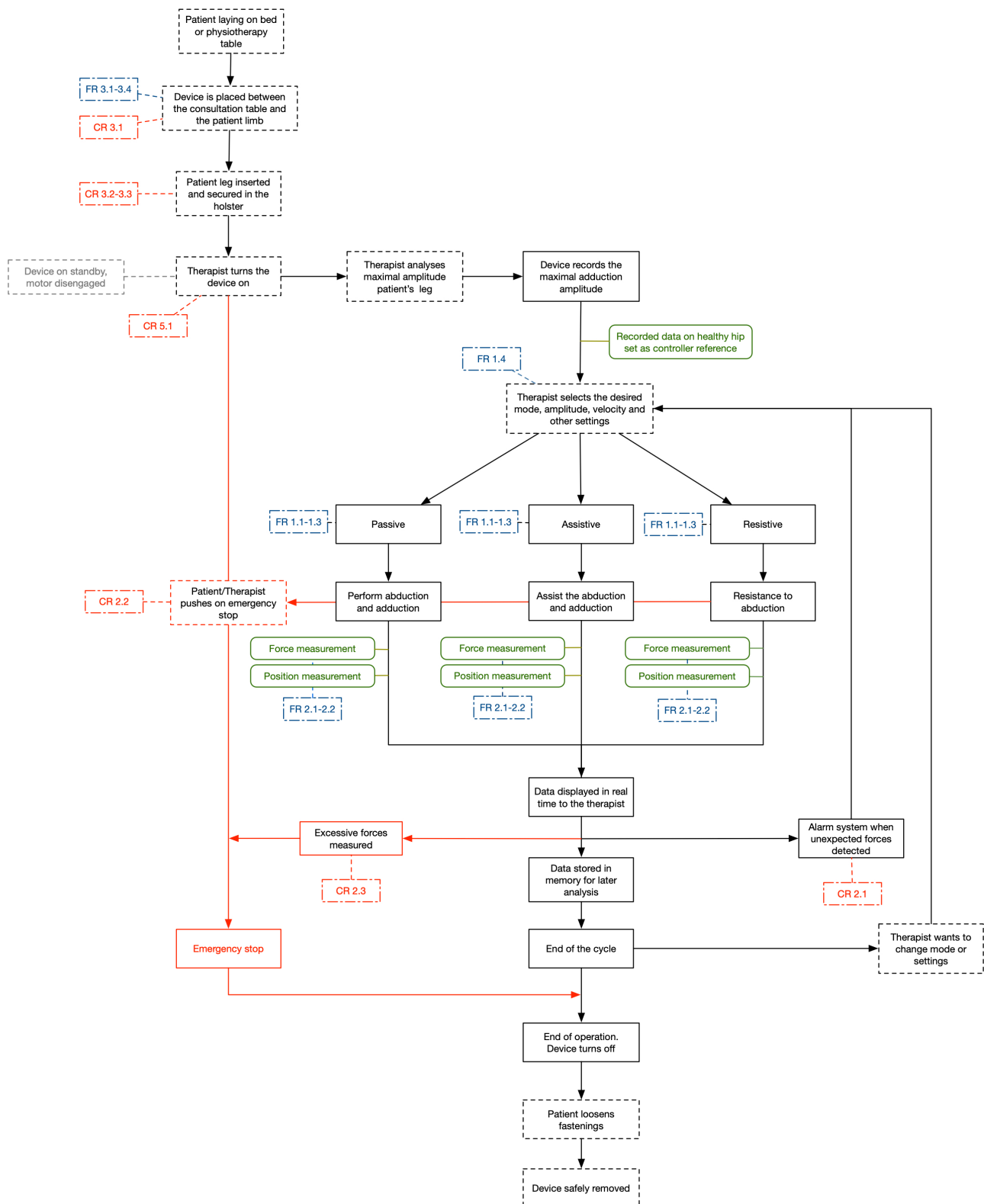


FIGURE 4.1: Functional diagram - Version 2 - June 2019

Chapter 5

Conceptual Design

The methodology followed to obtain the most optimal solution concerning the architecture of the actuation mechanism is detailed in this chapter.

First of all, we will briefly describe the results previously reached in the frame of the LMECA2855 course. On this basis, arises a variety of solutions, which will be presented and submitted through a selection process. To support this selection, a simulation in *Robotran* quantifies the efforts in the hip joint generated by the different solutions. Furthermore, a series of relevant criteria are defined to evaluate and classify the mechanisms in the most objective way. Finally, a classification table presents their scores regarding every criterion and serves as a basis for the final choice of the actuation system.

5.1 Methodology and results achieved in LMECA2855

In the framework of "*Mechanical design in biomedical engineering*" course, a first conceptual methodology was carried out, and an advanced design of M-Vitec was reached. After going through the main functions of the device, several ideas emerged among the team. Mostly, three primary mechanisms were considered for the actuation of the abduction and adduction: a crane-like mechanism, a scissors-like mechanism and a mechanism based on linear guides (Figure 5.1). A basic description of each mechanism is given below.

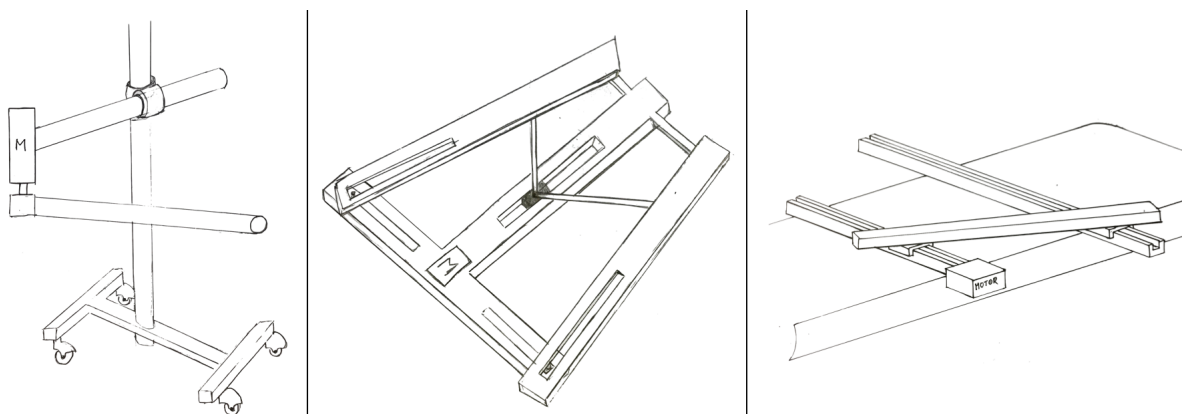


FIGURE 5.1: From left to right : crane-like mechanism, scissors-like mechanism, parallel linear guides mechanism [51]

The **crane-like mechanism** is an arm hanging above the bed, supporting the leg and performing abduction/adduction. The arm is actuated by a motor, situated on a trolley on the side of the bed. The rotating motor has to be precisely positioned above the instantaneous center of rotation of the hip.

Regarding the **scissors-like mechanism**, the patient sits on the device, with both legs fastened. They are independently actuated thanks to a linear actuator placed between the patient's legs. The rod that links the leg support and the actuator can be detached to work with one leg only. Again, the hip joint needs to be precisely aligned with the device rotation axis of the device.

The last solution is a mechanism based on two **parallel linear guides**, one of which is driven linearly. They support a third perpendicular rail, carrying a rotating disk where sits on the top, the leg support. The combination of two translations and one rotation allow the hip to guide the rotation without having to be aligned with any joint of the device.

Based on several criteria, such as ergonomics, mobility, safety, and price, these solutions were scored and ranked. The results of the classification indicated that the most adequate solution was the parallel linear guide mechanism. In fact, it stood out from the others, thanks to its ease and low duration of installation, great leg access from the physiotherapist and its reasonable estimated price. But above all, regarding the safety of the patient, the 3rd solution seemed to minimize the internal efforts in the weaken hip joint since it provided an extra degree of freedom, making the device to align with the hip and not the other way around.

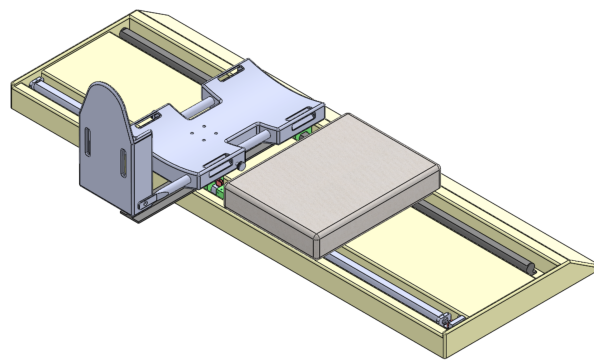


FIGURE 5.2: Isometric view of the Solidworks prototype resulting from LMECA2855 course [52]

After selecting the actuation mechanism, a detailed design was conceived with the CAD software *SolidWorks* [53]. The final prototype is illustrated in Figure 5.2 and 5.3. While the patient is laid down, the device is placed between the patient's lower limbs and the table/bed. The device is actuated through an electric motor connected to lead screw, which moves linearly the perpendicular rail. A support mobilizes the leg and keeps the patient's foot in an upright position, avoiding internal and external rotation. A comfortable cushion gives support for the healthy leg, but also provided extra stability to the device. Finally, a force sensor is integrated into the driven carriage and the motor is equipped with an encoder providing its angular position.

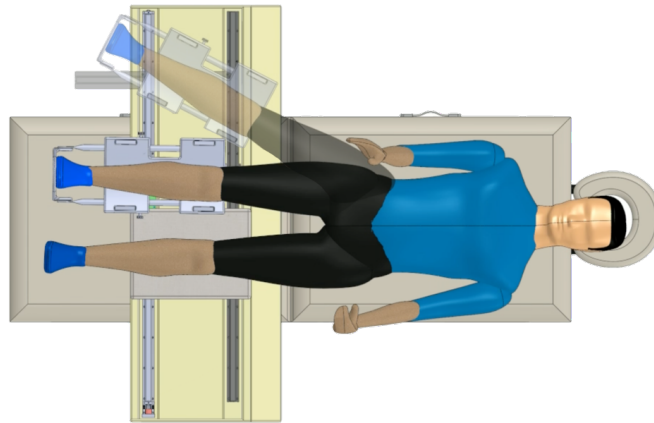


FIGURE 5.3: Overview of LMECA2855 prototype running with a patient [52]

5.2 Description of solutions

As detailed in the previous section, when starting this master's thesis, an advanced design of a solution has already been in progress. However, very soon, we realized that some aspects needed to be reconsidered in order to reach the most suitable device for this application. Hence, a backward step was taken to have a deeper reflection on the actuation mechanism.

The idea of having an extra degree of freedom in a planar mechanism was definitely a keeper since it allows the device to adjust according to the patient's leg, irrespective of its positioning and morphology. From that principle, we identified simple kinetic systems with three degrees of freedom that can generate a hip rotation in the coronal plane.

Nine different mechanical architectures were determined and considered in this conceptual phase. They are different combinations of revolute and prismatic joints and form three distinct kinematic families, as follow:

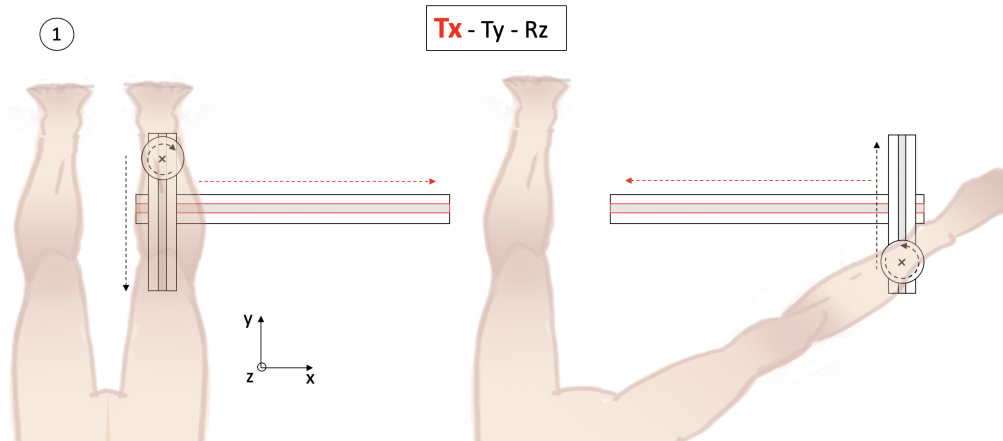
1. Two prismatic joints + One revolute joint: Drawings 1 to 6
2. One prismatic joint + Two revolute joints: Drawings 6 and 7
3. Three revolute joints: Drawing 8

In the following drawings, the linear guides correspond to the prismatic joints, while the revolute joints are represented by rotating disks. On the top of each drawing, an inset indicates the nature of the joints composing the mechanism (T stands for translation and R for rotation). As the drawings are a top view, the order of the joints, goes from the farthest to the closest, with respect to the patient's leg (highest element). The actuated degree of freedom corresponds to the red joint referenced in the insets and the red arrows in the drawings. Note these images are purely illustrative, so the dimensions may not respect exactly the same scale (leg representation inspired from source [54]).

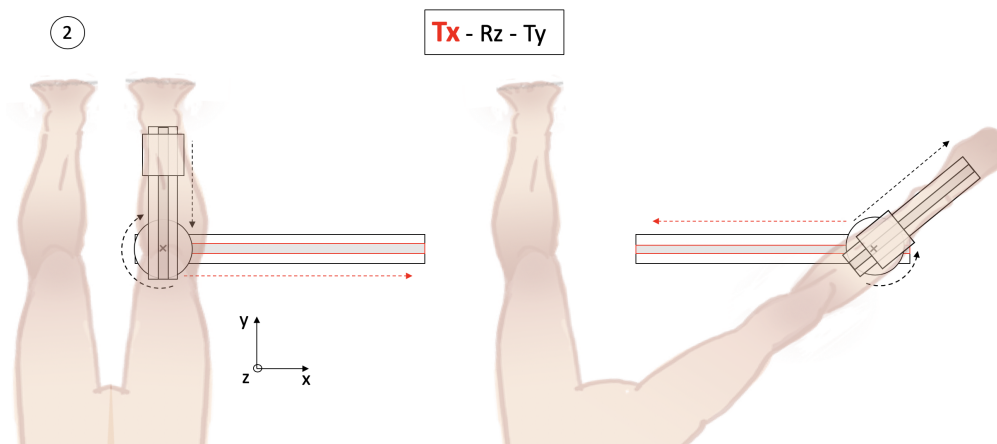
In addition, the arrangement of the components was established, emphasizing the device compactness and patient safety (singular configurations avoided) since they are intended to be positioned just under the patient's lower limb.

"Two prismatic joints + One revolute joint" solutions

The mechanisms belonging to the first group, are composed of two linear systems in both x and y directions, as well as, one rotary plate around the z-axis. Six different combinations were studied, all linearly actuated in the x-direction.



As you may notice, **solution 1** is the mechanism developed in the LMECA2855 course, but with only one rail along the x-direction. Indeed, the extra support rail provides more stability, but perfect parallelism between rails must be maintained to avoid considerable friction. While performing abduction/adduction movements, the shorter linear guide travels along the longest and fixed actuated linear guide.

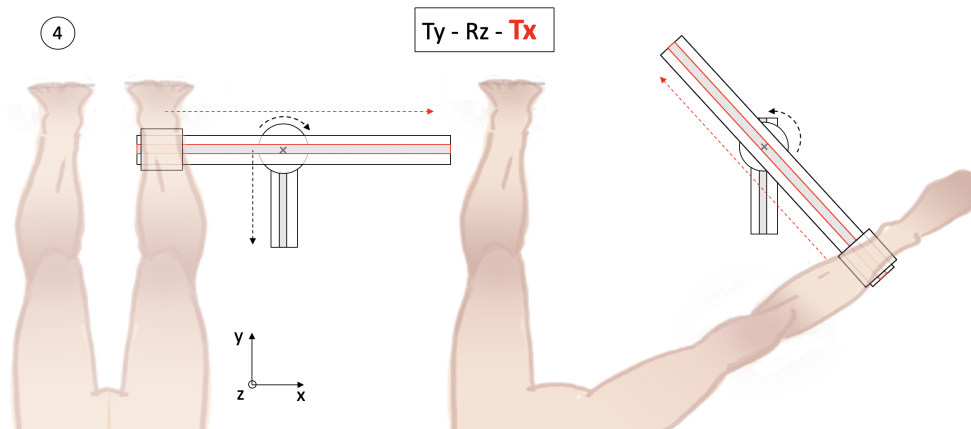
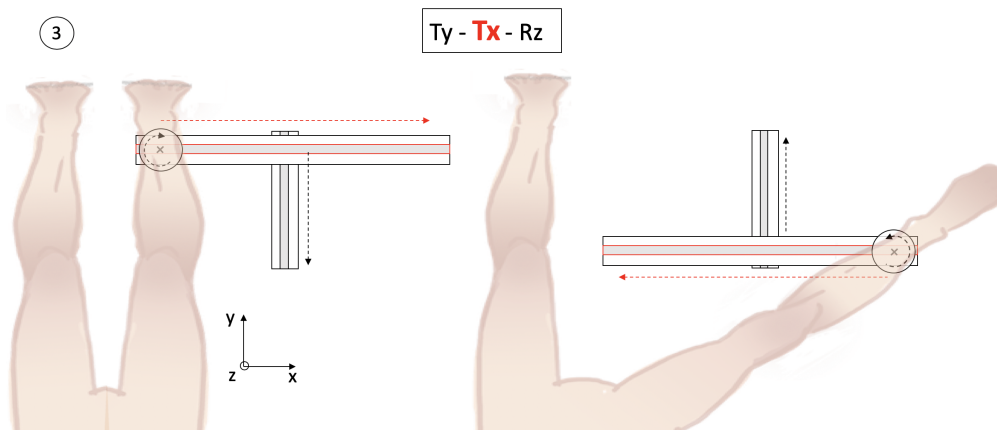


In **solution 2** the rotational degree of freedom occurs between the two linear guides. Thus, the shorter guide is always aligned to the patient's leg and the longer one keeps still. Again, the actuation is driven along the x-axis.

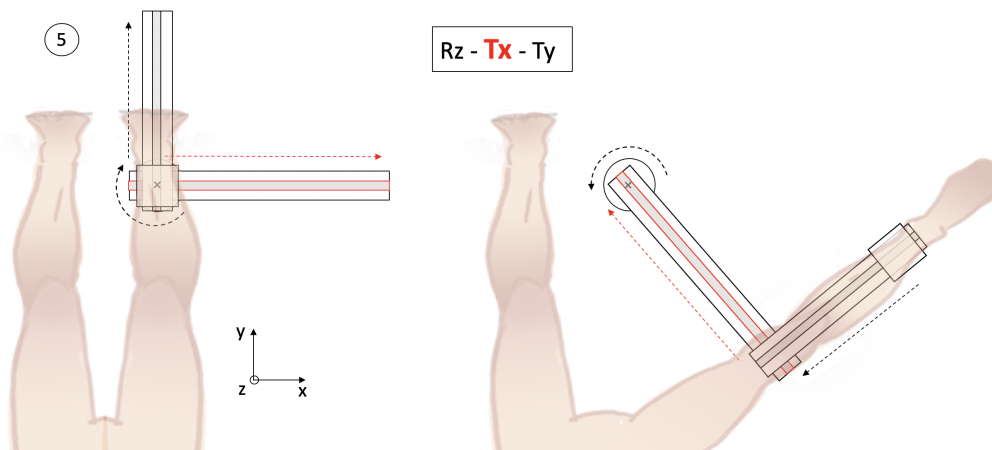
The **solution 3** is similar to the first mechanism, except that the x and y translations are reversed. This time, the longer and actuated rail moves back and forth in the y-direction while the shorter linear guide keeps still.

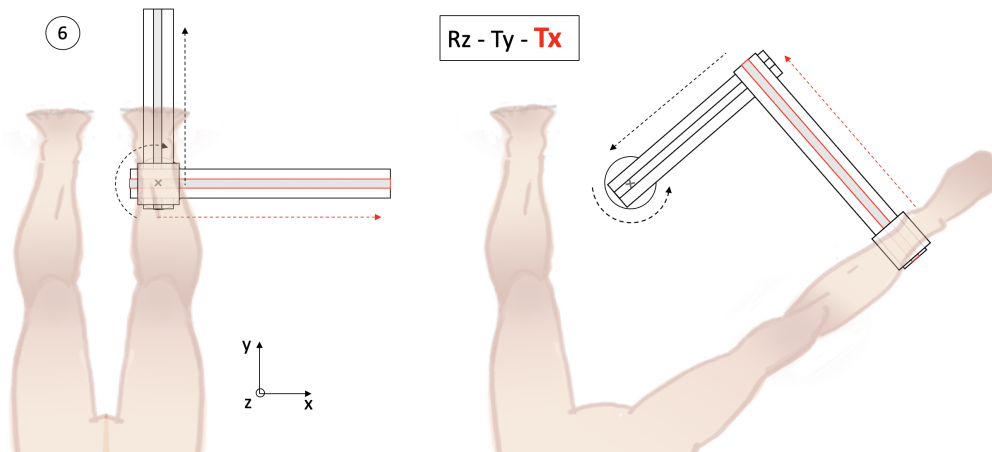
In **solution 4**, the revolute joint is placed between the two translations which keep the motorized guide always perpendicular to the patient's leg, illustrated drawing 4.

Thus, the longer rail pivots when the hip angle increases and decreases.



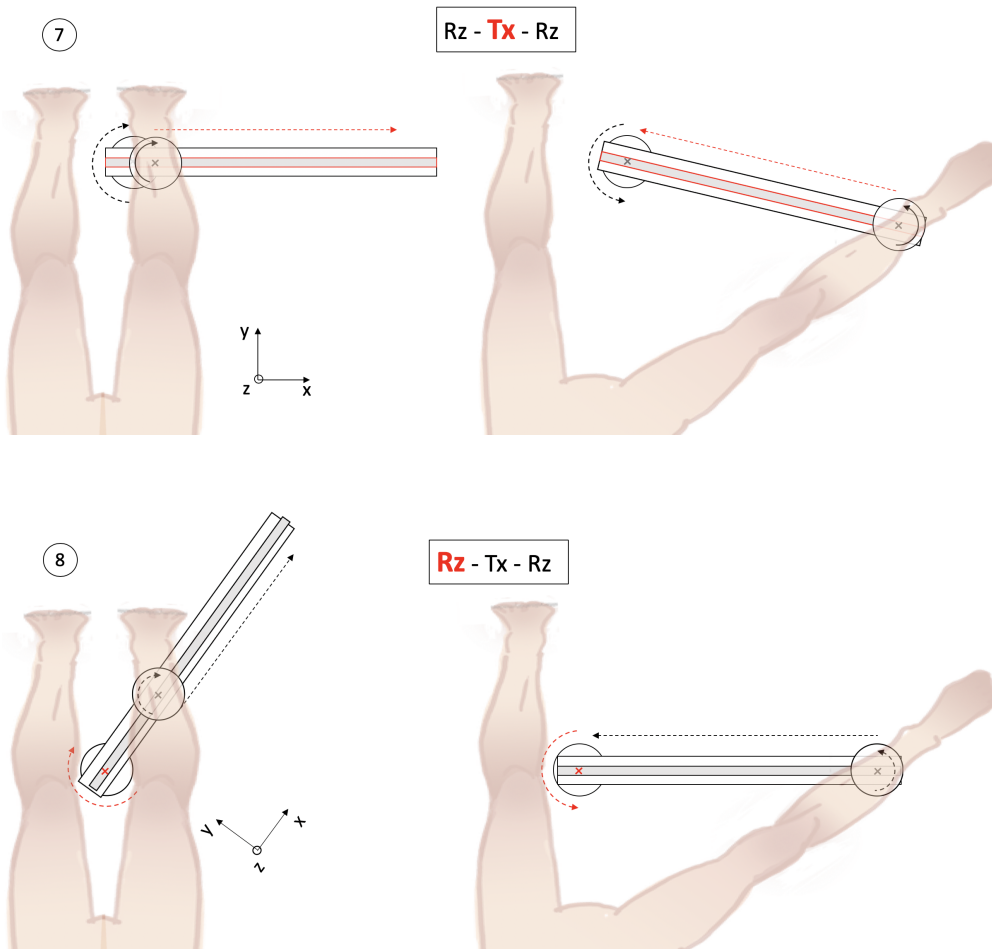
Solutions 5 and 6 are equipped with a rotary joint in the first position. Hence, as mechanism 4, the motor force is always acting perpendicularly to the leg. However, in contrast with the previous models, both mechanisms cause the displacement of the two linear guides when performing abduction/adduction movements.





"One prismatic joint + Two revolute joints" solutions

The systems from the second family are composed of two revolute joints and one linear guide. The two investigated mechanisms are similar, except for the transmission system and the initial angular position of the linear guide.



Solution 7, as all solutions discussed until now, the abduction/adduction is generated thanks to the actuation performed linearly, along the guide. During the movement, the guide will also pivot around the first revolute joint fixed to the device base

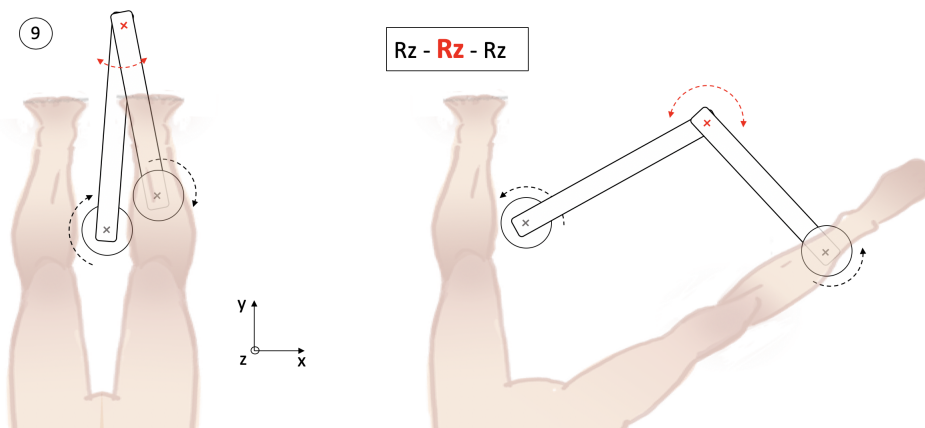
frame.

In contrast, **solution 8** is powered by a direct drive system in which the motor shaft is connected to the z-axis revolute joint between both legs. Thus, it induces the rotation of the linear guide around its fixed extremity (red cross), which force the carriage to travel linearly. As you can see in drawing 8, the linear guide starts the movement in an oblique position to avoid excessive efforts inside the hip.

"Three revolute joints" solution

The last type of mechanisms only contains one possible solution: **solution 9**. In fact, the studied architecture is comparable to a four-bar linkage formed by two non-extensible rods intrinsic to the mechanism, plus the leg itself and the fixed distance between the hip joint and the pivot next to the resting leg.

Again, the actuation is a direct drive which takes place in the joint connecting the two physical rods represented by a red cross in drawing 9. By increasing their angle, the legs are forced to move apart and the hip joint rotates accordingly.



In addition, we considered another variety of solutions which presented the same kinematics as the previously discussed mechanisms, but the device base is placed next to the bed and the leg is mobilized from above (similar to LMECA2855 crane-like solution). However, based on TBN insights and the models implemented in *Robotran* (explained further in this chapter), we realized that these alternatives were deviating from the established technical requirements, such as monopolizing too much space in the medical room and making transportation complicated.

Nevertheless, illustrations of some of these alternative solutions can be found in Appendix A, section A.

5.3 Robotran analysis

After the surgical procedure, the holding function of the tissues around the hip is severely perturbed. Thus, there are non-negligible chances of postoperative dislocations if the recommendations of the medical experts are not followed carefully. In general, it is crucial to not over-strain the hip joint, to allow faster and healthier tissues healing and limit the risk of dislocation.

The aim of this analysis is to quantify the internal efforts felt in the hip when performing the abduction/adduction movement with one specific actuation system. These forces are partly related to the inertial characteristics of the moving leg, but they are also caused by the device itself. Since each of the presented mechanism has its own way of interaction with the leg, they induce different efforts inside the hip joint. This investigation was performed thanks to *Robotran*, a software which enables to model and simulate multibody systems, developed by the Multibody research group of the Center for Research in Mechatronics at UCLouvain [55].

5.3.1 Robotran implementation

To identify the maximal forces generated in the hip joint, we evaluate the worst-case scenario by imposing the most intense trajectory within the range of amplitude, velocity, and acceleration defined in the technical requirements.

As the goal of the rehabilitation device is to provide isokinetic work (optimal for the tissue healing), the choice of hip angular trajectory was straightforward: a trapezoidal velocity profile, commonly used in robotics. Finally, these were the parameters which defined the angular position, velocity and acceleration profiles given to the hip joint:

- $\theta_{max} = 60^\circ$
- $\dot{\theta}_{max} = 60^\circ/s$
- $\ddot{\theta}_{max} = 90^\circ/s^2$
- Trapezoidal velocity profile

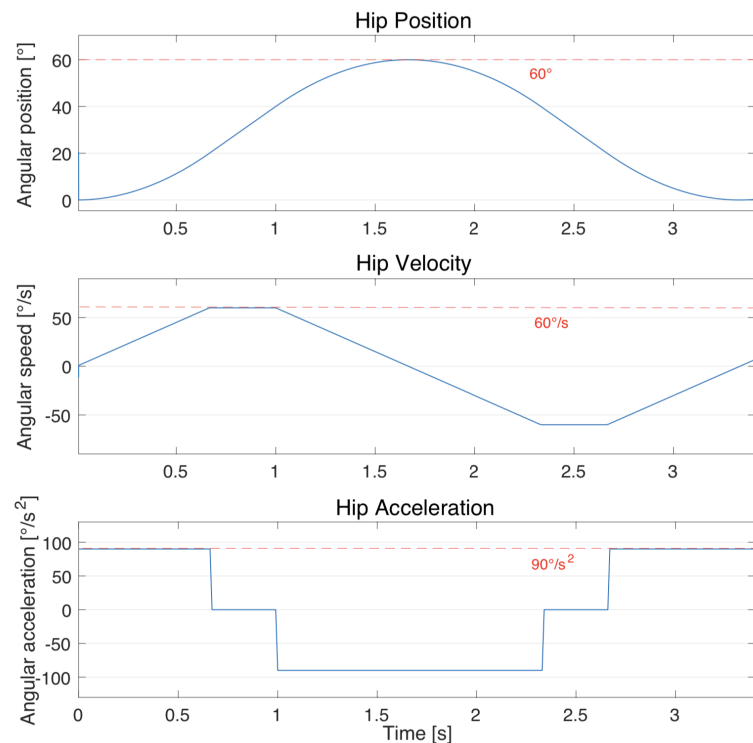


FIGURE 5.4: Angular hip trajectory used for the force analysis

Based on these parameters, we obtained the curves given in Figure 5.4 for a complete abduction/adduction cycle in approximately 3.4 s. The *Robotran* interface enables the

implementation of simplified models for multibody systems, perfectly adequate for the mechanisms described in the previous section.

At this point, all the components acting on the system must be characterized, by fixing the masses, the matrices of inertia and the center of mass positions, which needed a first reflection about dimensional requirements. In order to make this preliminary study feasible, some simplifications and assumptions were established.

General assumptions

- The dry and viscous friction within the mechanical components were not considered. All joints were modeled as perfect with no internal friction.
- The hip tissue resistance and hip joint stiffness were considered negligible compared to the other forces acting on the system, so they were not simulated. Thus, the hip joint was represented as a perfect revolute joint.
- The lower limb mass is supported in two punctual locations: just above the hip and within the leg support. In reality, the weight should also be spread along the posterior surface in contact with the bed under the hip. Hence, the load applied on the device was over-estimated. Moreover, the leg was assumed horizontally positioned (parallel to the bed).
- Since all the mechanisms are 2D and move in the x-y plane, only the diagonal component I_{zz} was taken into account in the matrix of inertia.

Modeling of the leg & its support

The calculations were based on the most demanding patient physical conditions, namely a person weighting 100 kg and 2 m high. Based on morphological tables of human body proportions, the dimensions and weight of each of the leg segments were determined [56][57].

First of all, as shown in Figure 5.5, the leg shape was simplified into three homogeneous bodies: two cylinders and one rectangular cuboid, standing for the thigh, the calf, and the foot respectively. The missing dimensions such as the cylinder's radius, were deduced by using the mean density of the human body: 1062 kg/m^3 [58].

Moreover, the leg support was modeled as a 2 kg rectangular cuboid, 300 mm long, 200 mm large and 50 mm high.

Finally, for the implementation in *Robotran*, the 3-body leg and its support were merged in one unique body since we assume the leg is a solid non-articulated body perfectly fixed to the support.

The final body weights 18.1 kg and has a center of mass in the average position of the four bodies weighted according to their masses.

The total moment of inertia was the sum of the individual moments of inertia plus a component due to the translation to the new center of mass (Huygens-Steiner theorem) [59].

Modeling of mechanical components

Only the major mechanical parts were considered in Robotran using two different geometries. The linear guides, the carriage, and rods were represented by rectangular cubes, while small flat-ten cylinders (disks) modeled the revolute joints.

All the components were assumed homogeneous masses, except the linear actuators. In fact, their weights, moments of inertia and centers of mass were readjusted due to the presence of a non-negligible motor (1.5 kg) in one of their extremities. The engines were placed in the extremity that experiences less motion, in order to minimize their inertial impact on the hip joint efforts. However, for some solutions, as the linear actuator is fixed, the motor placement will not affect final results.

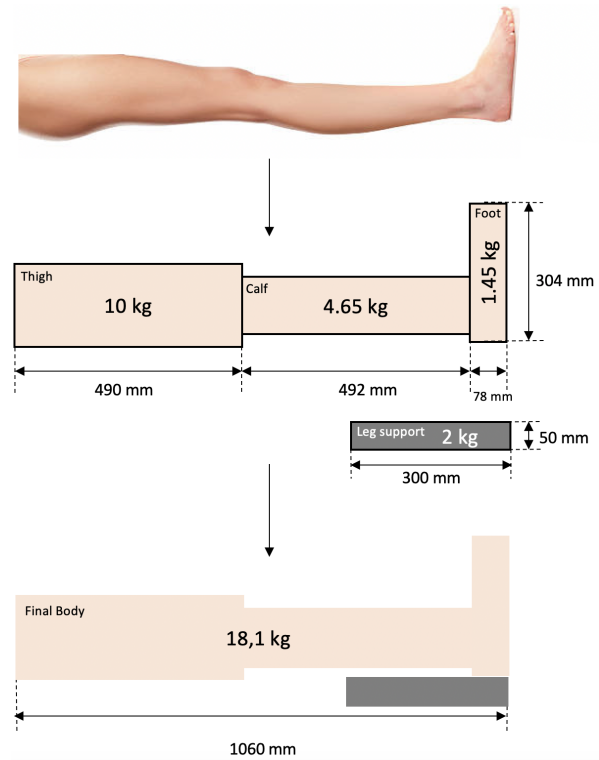


FIGURE 5.5: Leg modelling procedure

After investigating several existing linear guides, we realized that for the range of dimensions and force capacities needed, the overall mass density of the guides does not exceed 1000 kg/m^3 [60][61]. Thus, all the rectangular-shaped masses are based on that value. Their height and width were also inspired from real models.

To keep this analysis as reliable as possible, we imposed a constant distance (0.7 m) between the hip joint and the leg support for every mechanism. Based on that, we were able to deduce with geometrical relations (see section 7.1.1), the length of the linear guides in order to reach the desired maximal abduction amplitude of 60° .

5.3.2 Results & interpretation

Nine distinct models were integrated using the graphical editor of *Robotran*, called *MBSysPad*. For illustrative purposes, Figure 5.6 provides the final model and the 3D representation regarding solution 7. This model consists of four bodies and a fixed base, each having a configurable mass, matrix of inertia and center of mass. They are linked through six joints and one ball cut used to close the kinematic loop. The models of the other mechanisms are detailed in Appendix A, section A.

The *MATLAB* environment linked to the *Robotran's* interface, allows to manipulate and explore detailed kinematic and dynamic information of the implemented system.

As already reported, the first step was to impose the trapezoidal trajectory to the hip joint. The moving leg forces all mechanism to follow the given motion. Therefore, forces and torques are generated inside the joints of the closed system. By indicating the actuated joint, the software knows where is located the driven joint and simulates

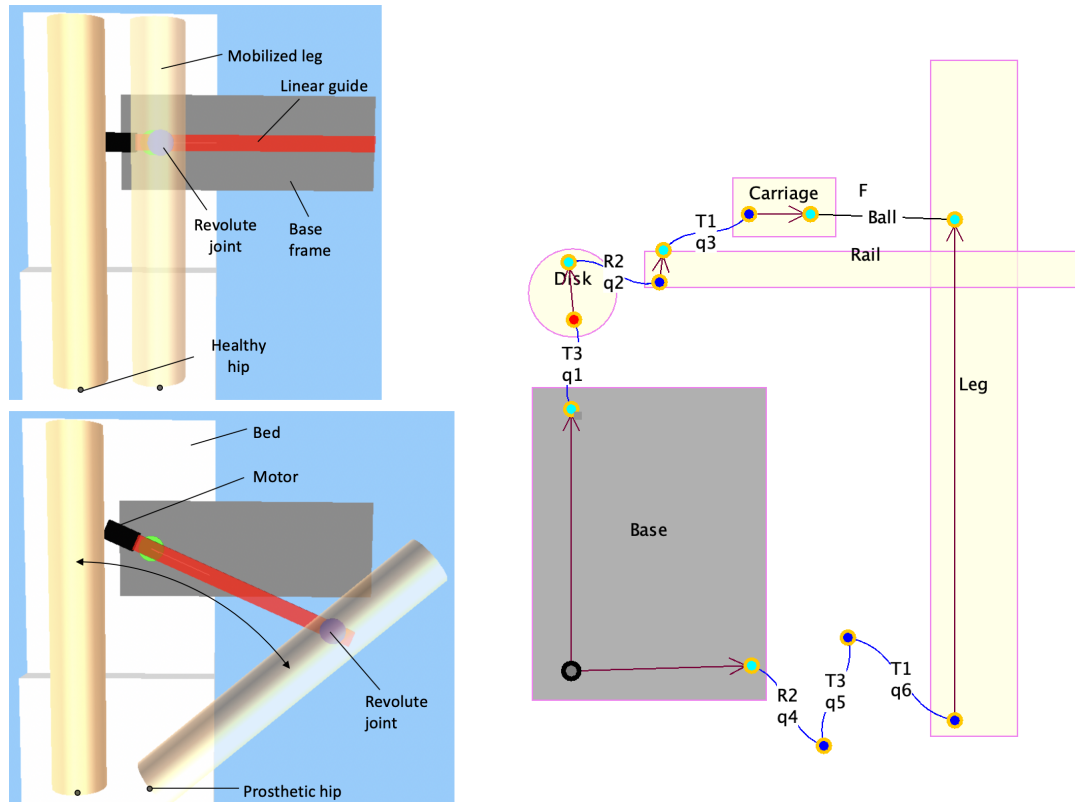


FIGURE 5.6: Robotran model and 3D representation of solution 5

its impact on the other joints.

Among a variety of available data given by *MATLAB*, we are mostly interested in the hip joint efforts while using the machine. These efforts can be decomposed in two components illustrated in Figure 5.7:

- F_a : Axial component, parallel longitudinally with the femur and leg axis.
- F_n : Normal component, perpendicular to the femoral segment.

Both components are measured according to an orthogonal frame fixed to the head of the prosthetic femur. Thus, the reference frame rotates with the leg. The normal and axial forces pointing towards the pelvis are considered positive values for further results.

We distinguish two kinds of phenomena causing stresses inside the hip joint.

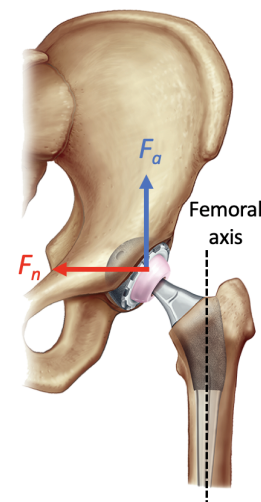


FIGURE 5.7: Force reference frame [62]

The first phenomenon is related to the natural physiology function of the hip. In fact, it is the mechanical joint (and the tissues around it) that connects the lower limb

with the rest of the body. In consequence, the hip has to retain the inertial effects whenever the leg is moving. In this case, the rotation of the leg induces centrifugal forces in the hip, which pulls the leg away from the hip (negative axial force). The maximum centrifugal force will happen when the higher velocity is reached and can easily be estimated as follow:

$$F_{c,max} = m * \omega_{max} * R = 18,1 * \frac{60 * \pi}{180} * 0.5 \simeq 9.5N$$

with, m : leg and support mass in [kg]

ω_{max} : maximum angular leg velocity in [rad/s]

R : distance between the leg's center of mass and the rotation axis in [m].

In *Robotran* simulation, the tissue surrounding the hip were not counted, so this force is fully applied to the hip joint. In reality, the centrifugal effect is partly supported by the tissues, so this force should be lower. In addition, this force varies from patient to patient, but should not exceed this value because the most critical patient was taken into account.

The second reason causing hip internal efforts is directly related to the device itself. The accelerated parts of the device, due to their inertial characteristics, will apply forces in the leg, which leads to forces inside the hip. Moreover, the direction of the motor force has a direct impact on the axial forces felt by the hip. For instance, in solution 4, 5 and 6, the actuation force is always perpendicular to the leg, so the device effect on the axial component is minimal throughout all the cycle (see Appendix A, section A, sol. 4, 5 and 6). Finally, all the devices allowing a rotational degree of freedom at the leg support (i.e. solutions 1, 3, 7 and 8), generate almost no normal forces in the hip.

The force profiles during an entire abduction/adduction cycle were obtained for all nine solutions. Hereby, will be presented the results for solution 7, but the results of the remaining solutions are detailed in Appendix A (section A), as well as, a summary table is provided at the end of this section.

Two axial force profiles are represented in Figure 5.8 for a complete cycle of 3.4 seconds. Note that a positive axial force means a compression of the hip joint while negative values represent its extension. The graph (a) takes into account all forces contributors while in the graph (b) leg centrifugal forces were subtracted. The second graph was useful to validate our estimation of the centrifugal force, but also to provide a clearer measure of the device influence on the axial component. The abduction movement can be divided into three phases:

- **[0 s - 0.6 s]:** As you can see in the graph (b), during the acceleration phase, the axial force is negative since the angle between the leg and the linear actuator becomes more and more acute, leading to a motor force that pulls the leg away. The hip joint extends.

- **[0.6 s - 1 s]:** In this interval, since friction is not considered, the plateau of maximal velocity is maintained with no motor force required. At this point, the only force acting on the system is the leg centrifugal force around -10 N shown in the graph (a). Thus, we notice a null force during this phase when leg centrifugal effects were ignored (graph (b)).

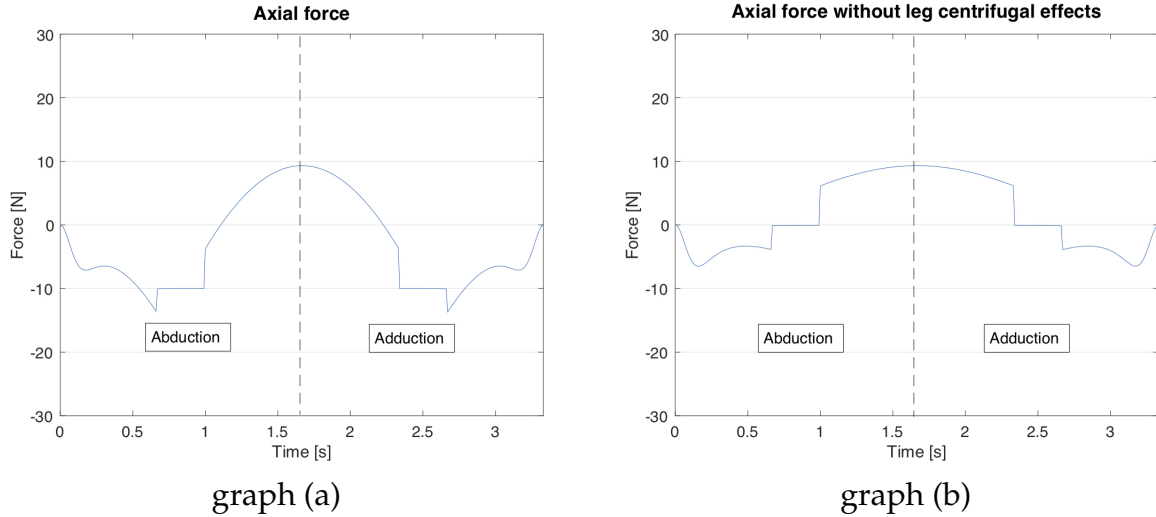


FIGURE 5.8: Axial forces (F_a) applied on the hip during the abduction/adduction cycle

- **[1 s - 1.7 s]:** In the last phase of abduction, the leg is decelerated by the motor force, which provokes a compression of the hip rising 10 N at the end of the abduction. In the graph (a), we observed a bigger variation of the axial force (from -10 N to 10 N) since the hip goes from extension to compression. Whereas, in the graph (b) the variation is softer (from 0 N to 10 N).
- **[1.7 s - 3.4 s]:** The adduction force profile follows exactly the same pattern as for abduction, but reversed because the mechanism starts at a 60° amplitude and comes back to the initial position with exactly the same trajectory characteristics.

The normal forces acting on the hip are plotted in Figure 5.9. For this solution, they are very low (between -0.5 and 0.5 N), which avoids any risk of dislocation. They are directly proportional to the leg acceleration, thus are likely caused by the inertia of the mechanical parts of the device. While for this solution they are minimal, in some others they cannot be neglected at all (i.e. solution 6, Table 5.1).

The last graph (Figure 5.10) illustrates the motor force at the transmission system's output in the direction of the linear guide, required to follow the abduction/adduction cycle given in Figure 5.4. As expected, the force curve is closely related to the leg acceleration profile. The small difference is explained by the non-perpendicular angle between the leg and the linear guide. The maximal force (-17.5 N at 1.6 s) is reached at an amplitude of 60° (Figure 5.6, bottom left), since the lowest angle between the leg and the linear guide is reached. We cannot forget that the mechanical losses inside

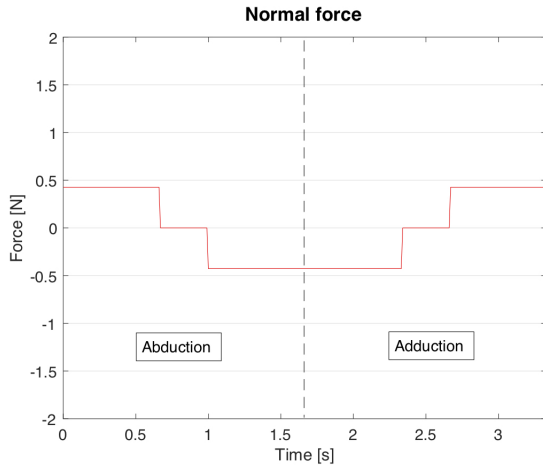


FIGURE 5.9: Normal force (F_n) applied on the hip during the abduction/adduction cycle

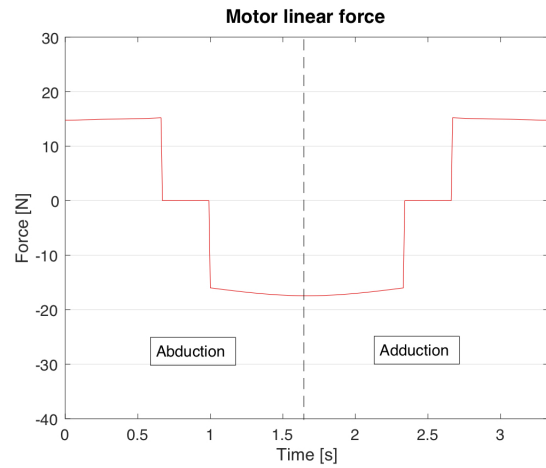


FIGURE 5.10: Motor force (F_m) required to perform the abduction/adduction cycle

the mechanism are not accounted, so in practice the motor force should be higher.

Table 5.1 gathers the extreme values for both axial and normal forces obtained for all the solutions. $F_{a,max}$ and $F_{a,min}$ stand for hip axial compression and extension respectively. While, $F_{n,max}$ and $F_{n,min}$ correspond to internal and external normal force respectively. As a reminder, the indicated values were evaluated according to the frame shown in Figure 5.7. In addition, the motor requirements are reported in the two last lines of the table. For the seven first solutions, the maximal force $F_{m,max}$ required from the linear guide is indicated. For the last two solutions, since they are actuated in rotation, the maximal torque $T_{m,max}$ was computed.

solutions	1	2	3	4	5	6	7	8	9
$F_{a,max}$ [N]	25.5	0	31	9.4	0	0	9.3	4.3	12.9
$F_{a,min}$ [N]	-15.2	-10	-18.7	-13.4	-10	-10.2	-13.7	-23.9	-18.5
$F_{n,max}$ [N]	0.4	11.8	0.4	4.4	2.8	2.2	0.4	0.4	0.4
$F_{n,min}$ [N]	-0.4	-5.6	-0.4	-0.2	-18.3	-20.7	-0.4	-0.4	-0.4
$F_{m,max}$ [N]	30	27.1	34.2	17.1	33.5	35	17.4	-	-
$T_{m,max}$ [N.m]	-	-	-	-	-	-	-	13.7	7.8

TABLE 5.1: Summary of *Robotran* analysis results

5.4 Solution classification

In this part, all the different mechanisms will be compared through a number of criteria, in order to identify the most suitable solution for this specific application. Each of the criteria will be described, by justifying their importance and how they were evaluated. Then, all the criteria are assembled in one unique table providing a classification of the solutions. This table will reveal the most adequate actuating mechanism and the one considered for the rest of this project.

5.4.1 Criteria description

Since we are in a preliminary phase of the design, for some of the criteria, it was difficult to assign a numerical value, so the systems were scored based on a relative comparison to a reference solution, solution 1 (sol.1) from LMECA2855.

Throughout the criteria selection, we notice that some criteria were intrinsically correlated, so they were regrouped in one unique criterion to avoid over-counting and keep the classification as objective as possible. Finally, only three different categories of criteria were considered: price, bulkiness, and safety. In each of the categories, several relevant criteria were identified.

Price

Financial limitations lead us to take into account the cost of the prototype for this classification.

- **Motorized linear guide:** The component impacting the most the final cost is the use of linear guides with an integrated drive system, whether a toothed belt or a screw. This criterion was simply quantified according to the number of actuated linear guides.
- **Non-motorized linear guide:** Needed for some solutions, although they are cheaper, they also have a substantial influence on the cost. The score was given in a similar way as the previous criteria.

The rest of the components were not considered because, either they are not significantly expensive, or they are necessary for all the mechanisms.

Bulkiness

The following set of criteria covers all factors that could influence the comfort of the patient and the workload for the therapist.

- **Space swept by the device compared to sol.1:**
This criterion examines the bulkiness generated by the device while operating in the clinical environment. An estimation of the device's width and length were obtained by computing geometrically the minimal lengths of the main components needed to reach a 60° angular hip abduction and considering the

dimensional assumptions detailed in section 5.3.1. Moreover, the 3D representations created with *Robotran* supported our estimation. The smaller is the area swept by the moving components, the less bulky is the device for the patient and physiotherapist.

A mechanism was considered comparable to sol.1, when the percentage difference between its occupied area and the sol.1's area was less than 10%.

- **Height of the device compared to sol.1:**

As claimed in the requirements, the device must be placed between the bed / table and the patient. Therefore, its height is a general factor to minimize to improve bulkiness. In fact, it is particularly important to maintain the leg as straight as possible in the horizontal plane (15° maximum in flexion) to avoid the risk of harming the weakened joint. The total height was estimated only taking into account main components using the heights detailed in Appendix E, table E.1. A mechanism was considered comparable to sol.1, when the percentage difference between its height and the sol.1's height was less than 10%.

- **Weight of the device compared to sol.1:**

The weight of the device will affect the ease to transport, move and place the system in its working environment which represents an important factor for the therapist's daily work. Again, only the main components were considered in the estimation of the total weight and the values taken are the same used in *Robotran* given in Appendix E, Table E.1.

A mechanism was considered comparable to sol.1, when the percentage difference between its weight and the sol.1's weight was less than 10%.

- **Compactness for transportation:**

The device must be mobile so it can be easily moved by the therapists inside a medical center, but also in between medical centers and patient's home (possible to insert in a normal car trunk). This criterion was estimated based on the possibility to fold up (ex: align two rails) the device for easier transportation. The dimensional characteristics were not considered in this criteria because they were already accounted in, previously the "Space swept" criterion.

Safety

The device will act in a medical environment where safety is the first priority. The following criteria concern all the factors that could put in danger the user of the device but also the medical practitioner.

- **Maximum/Minimum hip axial force:**

This criterion takes into account the axial forces felt by the hip joint extracted from the *Robotran* analysis. As mentioned before, it is crucial to limit as possible the efforts inside the hip, especially shortly after the surgery. Since we are trying to minimize both compression and extension of the hip longitudinally to the leg, it was judicious to split normal forces into two different criteria in the classification table.

- **Maximum/Minimum hip normal force:**

Moreover, excessive normal forces are surely not recommended for a weakened hip joint. As the axial forces, we considered the most negative and the most positive normal force, since both are undesired for the patient. Again, the numerical values were obtained thanks to the simulations in *Robotran*.

- **Risk of collision with the patient compared to sol.1:**

Every system is based on different arrangements of linear and rotational components. In fact, in order to obtain the same leg trajectory, they move differently in the space. This criterion visually assesses the systems according to their proximity to the human body. The closer the parts move with respect to the body, the higher is the risk of harming the user in case of unexpected behavior of the device.

- **Risk of collision with an external body compared to sol.1:**

By contrast, we should also limit the space invaded by the moving elements outside the bed or table, to avoid any risk of collision with external elements. Besides, usually the available space around the patient in a clinical room is restricted.

- **Risk of trapping compared to sol.1:**

This criterion estimates how alike the device could be blocked mechanically. The jamming could be caused by patient's clothes, bed sheets or other external objects. In any case, this risk must be minimized in order to avoid any accident. The systems were classified, based on the number of joints and their exposure. In fact, regarding this criterion and the two previous, a hard shell covering all mechanical parts would avoid the risk of collision, but since we can not guarantee the compatibility of such protection for all the solutions, we were forced to maintain these criteria.

- **Mechanical resistance compared to sol.1:**

Mechanical resistance is a source of danger to the user since it provokes an unexpected cinematic of the device which could even cause a blockage of the system. In fact, these events could lead to violent movements that could be very dangerous for the patient. The resistance was estimated based on the nature of the joints composing each of the mechanisms. We assumed linear rails present higher friction than rotational joints.

5.4.2 Scoring sheets

The following sheets (Figures 5.11 and 5.12) gather the results of the classification process for the actuation solutions.

The scoring metrics are divided into different levels, going from 0% to 100%. A percentage score is assigned to each criterion. Then, an average score for each category is computed, called category score.

The importance given for each group of criteria was defined with the help of M.Thanh Binh Nguyễn by setting a weighting factor out of 100% for each category, shown at the bottom of Figure 5.12.

Finally, the final score of each solution is the sum of the category scores multiplied by its respective weight. Thus, the total score is expressed in %, 100% being the maximum score possible.

5.4.3 Selected actuation architecture

By observing the final results of scoring sheets, we notice that solution 7, 8 and 9 clearly stand out from the others. In fact, both direct drive solutions belong to this final group with an identical score of 71%. Solution 7 positions itself at the top of the ranking with a high score of 80%. Therefore, solution 7 was the selected solution.

This choice was congruent with the expert opinion of TBN and supported by the results obtained in the *Robotran* analysis. By taking a closer look to Table 5.1, we noticed that despite their good rating, solutions 8 and 9 induce a very strong axial extension of the hip (-23.9 and -18.5 N) which is conducive to a joint luxation. Moreover, the motor force at the output of the transmission system, necessary to drive solution 7, is approximately half compared to most of the linear actuated systems.

As already presented in Figure 5.6, the selected actuation solution is composed of 2 revolute joints and one prismatic joint which is a long motorized rail. Hence, this system is a three degrees of freedom planar mechanism installed on a bed or a therapy table which fixes the leg from below.

				<u>Scoring sheet - 1</u>								
				2 Translations + 1 Rotation						1 Translation + 2 Rotations		3 Rotations
Criteria category	Criteria	Metric		Sol 1	Sol 2	Sol 3	Sol 4	Sol 5	Sol 6	Sol 7	Sol 8	Sol 9
Price	Motorized linear guide	0	100%	0%	0%	0%	0%	0%	0%	0%	100%	100%
		1	0%									
	Non-motorized linear guide	0	100%	0%	0%	0%	0%	0%	0%	100%	0%	100%
		1	0%									
			Average	0%	0%	0%	0%	0%	0%	50%	50%	100%
Bulkiness	Space swept by the device compared to Sol 1	Smaller	100%	50%	0%	50%	0%	0%	0%	50%	0%	0%
		Comparable	50%									
		Bigger	0%									
	Height compared to Sol 1	Lower	100%	50%	50%	50%	50%	50%	50%	100%	100%	100%
		Comparable	50%									
		Higher	0%									
	Weight compared to Sol 1	Lighter	100%	50%	50%	50%	50%	50%	50%	100%	100%	100%
		Comparable	50%									
		Eavier	0%									
	Compactability for transportation compared to Sol 1	Better	100%	50%	100%	50%	100%	50%	50%	100%	100%	100%
		Comparable	50%									
		Worst	0%									
			Average	50%	50%	50%	50%	38%	38%	88%	75%	75%
Safety	Maximum axial hip force	0 - 5 N	100%	0%	100%	0%	75%	100%	100%	75%	100%	50%
		5.1 - 10 N	75%									
		10.1 - 15 N	50%									
		15.1 - 20 N	25%									
		> 20 N	0%									
	Minimum axial hip force (absolute value)	0 - 5 N	100%	25%	75%	25%	50%	75%	50%	50%	0%	25%
		5.1 - 10 N	75%									
		10.1 - 15 N	50%									
		15.1 - 20 N	25%									
		> 20 N	0%									

FIGURE 5.11

				Scoring sheet - 2									
				2 Translations + 1 Rotation						1 Translation + 2 Rotations		3 Rotations	
				Sol 1	Sol 2	Sol 3	Sol 4	Sol 5	Sol 6	Sol 7	Sol 8	Sol 9	
Criteria category	Criteria	Metric											
Safety	Maximum normal hip force	0 - 5 N	100%	100%	50%	100%	100%	100%	100%	100%	100%		
		5.1 - 10 N	75%										
		10.1 - 15 N	50%										
		15.1 - 20 N	25%										
		> 20 N	0%										
	Minimum normal hip force	0 - 5 N	100%	100%	75%	100%	100%	25%	0%	100%	100%	100%	
		5.1 - 10 N	75%										
		10.1 - 15 N	50%										
		15.1 - 20 N	25%										
		> 20 N	0%										
	Risk of collision with the patient compared to Sol 1	Less	100%	50%	50%	0%	0%	0%	50%	100%	50%	100%	
		Comparable	50%										
		Higher	0%										
	Risk of collision with an external body compared to Sol 1	Lower	100%	50%	50%	50%	0%	50%	0%	50%	50%	0%	
		Comparable	50%										
		Higher	0%										
	Risk of trapping compared to Sol 1	Less	100%	50%	50%	50%	0%	0%	0%	100%	100%	0%	
		Comparable	50%										
		Bigger	0%										
	Mechanical resistance compared to Sol 1	Better	100%	50%	50%	50%	50%	50%	50%	100%	100%	100%	
		Comparable	50%										
		Worst	0%										
			Average	53%	63%	47%	47%	50%	44%	84%	75%	59%	
	Criteria category		Weight	Sol 1	Sol 2	Sol 3	Sol 4	Sol 5	Sol 6	Sol 7	Sol 8	Sol 9	
Price		15%	0%	0%	0%	0%	0%	0%	50%	50%	100%		
Bulkiness		35%	50%	50%	50%	50%	38%	38%	88%	75%	75%		
Safety		50%	53%	63%	47%	47%	50%	44%	84%	75%	59%		
		Total	44%	49%	41%	41%	38%	35%	80%	71%	71%		

FIGURE 5.12

Chapter 6

Actuation system dimensioning

After selecting the global architecture of the actuation system, we could proceed to the detailed dimensioning and choice of the linear actuation system. It is mainly composed by a guide rail equipped with an integrated engine and transmission system, which transforms rotatory motion in a straight line motion of a carriage.

6.1 Actuator technical requirements

Before looking deeper into the different kinds of linear transmission solutions and the available technologies in the market, complete technical specifications must be established regarding several characteristics of the linear actuator.

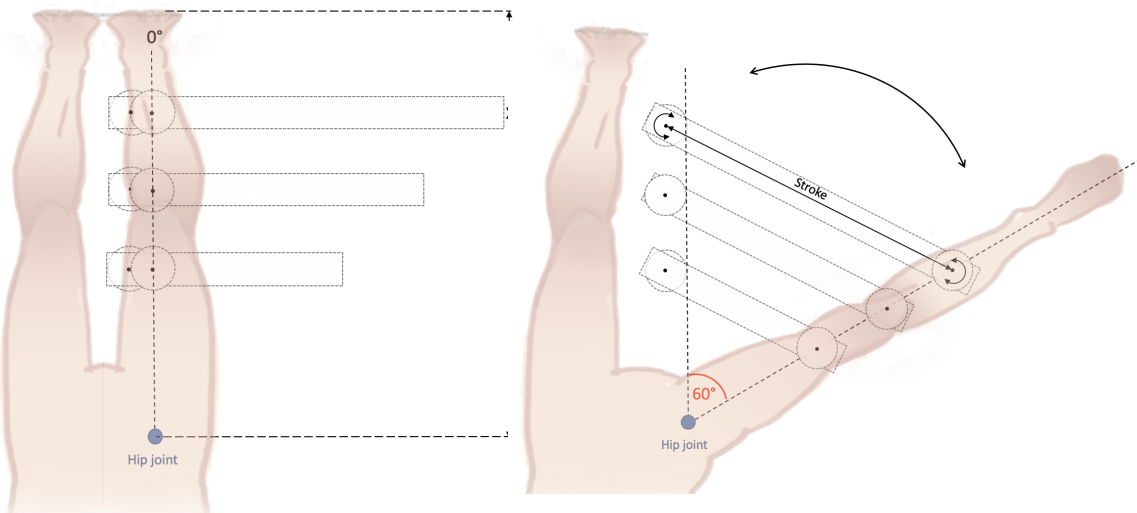
6.1.1 Dimensional requirements

The first step for its dimensioning was to derive the needed rail's size (stroke, height, and width), allowing to reach the trajectory profiles set in the technical requirements, for every body's morphology. Additionally, a weight limit should be fixed.

Stroke

Our starting point was to set the distance between the hip joint and the contact point between the patient and the device (leg support), D_{sup} . In fact, this value has a direct impact on several aspects:

1. **Rail length:** as you can see in Figure 6.1, the farther the rail is positioned from the hip joint, the greater will be the required stroke to perform the desired 60° abduction amplitude. Thus, ultimately it affects the overall width and weight. Consequently, a bigger guide leads to a bulkier device, less mobile.
2. **Carriage velocity & acceleration:** naturally, for a longer rail, the carriage needs to travel at higher velocity and acceleration to generate the same angular leg trajectory.
3. **Hip efforts:** due to the leverage effect, the greater is D_{sup} , the lower will be the required motor force to induce the same torque around the hip. Consequently, a lower motor force means less axial and normal forces sensed inside the hip joint.

FIGURE 6.1: Impact of D_{sup} on the linear carriage stroke

The two first elements drive us to minimize D_{sup} , to improve the device's compactness and reduce the weight of the rail. In contrast, increasing D_{sup} , lowers the required motor force and thus emphasizes patient safety by reducing the undesirable efforts inside the hip joint. Therefore, a trade-off must be established juggling between a reasonable device size, low stresses in the hip and the compatibility with all leg shapes. To make an informed choice of D_{sup} parameter, a sensibility assessment was conducted.

Thanks to *Robotran* models, we were able to measure precisely the influence of the leg support position, on the dynamics of the system, namely on axial and normal hip forces. The patient's physical characteristics and trajectory parameters were kept the same as in the previous analysis.

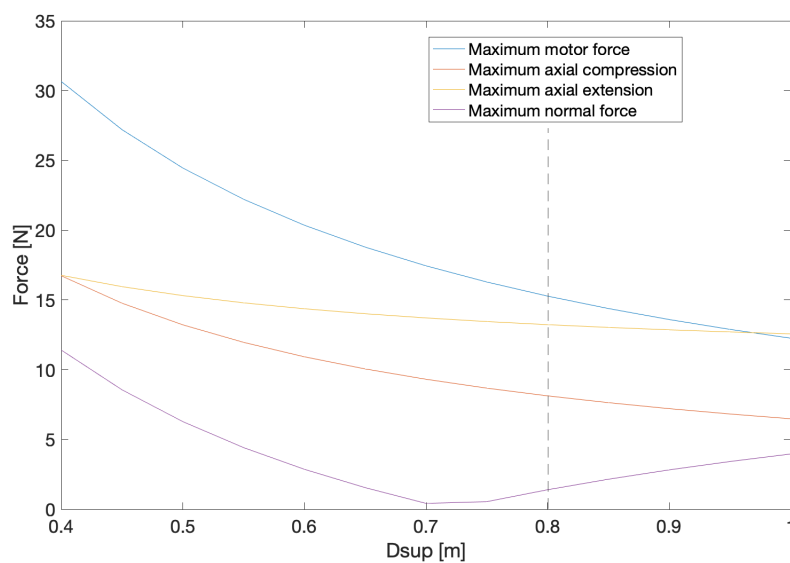
FIGURE 6.2: Evolution of the hip efforts and the required motor force with respect D_{sup}

Figure 6.2, presents the evolution of the maximal values of different forces while D_{sup} varies from 0.4 m to 1 m. As D_{sup} increases, axial compression, and extension reduce continuously (yellow and orange lines). As expected, the actuation force (blue line) also decreases while D_{sup} gets bigger, achieving an impressively low force of 12.5 N at the highest D_{sup} . In addition, the maximum normal force (purple line) decline until D_{sup} equals to 0.7 m, but then starts to raise for the higher values of D_{sup} .

We have to keep in mind that these profiles are based on a 2 m patient having a lower limb measuring 1.06 m long. One thing is certain, for every leg morphology, the distance between the linear guide and the foot's extremity will always be constant (D_{foot} in Figure 6.1). Looking again at Figure 6.2, fixing D_{sup} to 0.8 m seems to be a balanced trade-off, leaving 0.26 m between the foot and the rail. In fact, for D_{sup} higher than 0.8 m, the axial forces nearly decrease and normal force starts to considerably manifest. However, for shorter patients (smaller limb) the linear guide will push closer to the patient's hip, leading to:

- A shorter lever arm, resulting in a higher motor force to obtain the same torque around the hip.
- A leg with less weight and moment of inertia which decreases the forces needed to accelerate the leg.
- A leg center of mass closer to the hip joint and the reduced weight inducing lower centrifugal forces.

Since the physical characteristics of the patient are considerably involved in the dynamics of this system, it was appropriate to validate this parameter choice for other patient morphology. For that purpose, we compute the hip efforts for medium and small size patients, where the leg support and the foot are separated by 0.26 m. The results are reported in Table 6.1. Note that a weight of 100 kg was considered independently of patient size, which leads to overestimated results.

Patient morphology	Small	Medium	Large
Height	1.3 m	1.65 m	2 m
Weight	100kg	100kg	100 kg
D_{sup}	0.49 m	0.68 m	0.8 m
Max. axial compression	7.1 N	7.8 N	8.1 N
Max. axial extension	10.2 N	12 N	12.5 N
Max. normal force	1 N	0.3 N	1.5 N
Max. motor force	12.6 N	14.3 N	15.3 N

TABLE 6.1: Force comparison table for different patient sizes

After all, we notice that the largest patient will suffer higher efforts within the hip, as well as a bigger motor force is required to perform the movement. Thus, fixing D_{sup}

to 26 cm appeared to be a good choice.

Finally, thanks to trigonometric relationships (detailed section 7.1.1), we could finally deduce the stroke (carriage displacement over the rail) allowing a 0° to 60° abduction for our range of patient's size. For a $D_{foot} = 0.26$ m, a stroke $S = 0.8$ m was sufficient and leads to a rail length of about 1 m.

Height, width & weight

Regarding the rail's height, the lower the better, since the total height of the device needs to be reduced to minimize hip flexion. No real limitations are necessary for the rail's width, although the more compact is the rail, the less the device will be bulky under the patient's leg. Thus, the limit for the height and width is set to 8 cm and 10 cm respectively.

We also favored lighter linear systems to ease the device's mobility, expecting an actuation system not exceeding 6 kg (motor included).

6.1.2 Drive system performance

Linear velocity & acceleration

Every linear actuator is limited in velocity, acceleration and output force. The drive system capacities must cover the most extreme conditions of force and trajectory. After defining D_{sup} , we have easily derived the maximum speed and acceleration the carriage will have to travel, respectively 0.83 m/s and 1.34 m/s².

Motor force

The output force is the linear force needed at the output of the transmission system to move the leg. *Robotran* computes a maximal value of 15.3 N but this estimation does not take into account any friction, such as the tissue resistance around the hip. Therefore, it seems reasonable to increase this value to 50 N. Note that the maximal motor torque must also consider the mechanical friction generated inside the guidance system.

Accuracy

Regarding position accuracy, the specifications imposed an angular accuracy of 2° for hip abduction. This corresponds to a fairly low position accuracy within the guide rail, which is ensured by the majority of linear drive systems.

6.1.3 Carriage load & moment capacity

The last important characteristic that must be defined is the maximal loads and moments that are withstood by the carriage. Among the three forces and three moments illustrated in Figure 6.3, only the vertical load F_z is non-negligible.

In fact, the leg and its fixation are placed over the carriage which supports part of their weights. Taking the heaviest patient and a 2kg leg fixation, we an estimated

$F_{z,max}$ load not higher than 105 N (just half weight of the leg supported by the device). To be sure, we consider a maximal vertical load of 300 N (2.85 as safety factor) to also handle the eventual downward leg movements.

Finally, the research of a suitable linear actuator was based on the following specifications:

Specifications	
Stroke	800 mm
Max. rail height	80 mm
Max. rail width	10 mm
Max. weight	6 kg
Output force	50 N
Max. velocity	0.85 m/s
Max. acceleration	1.35 m/s ²
$F_{z,max}$	300 N

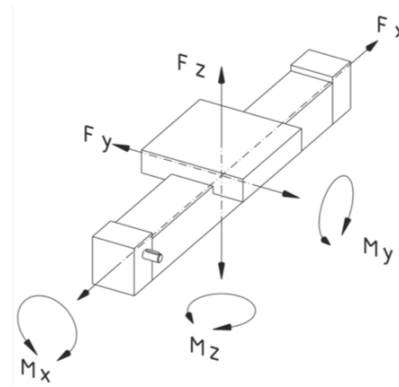


FIGURE 6.3: Carriage loads and moments [60]

6.2 Overview of the linear motion systems

There are a variety of ways to convert rotational motion from an electrical motor into useful linear motion. Among them, according to our requirements and device's environment, three transmission systems stood out: the rack and pinion mechanism, the ball screw assembly and the toothed belt drive.

Rack & pinion transmission

This mechanism is composed of a circular gear and a rack, which transform rotational motion into linear motion. When the rack is fixed, the rotation of the pinion forces it to move along the rack.

Linear rails equipped with this type of drive systems exist, such as the SQ-ZST model by RK Rose+Krieger company (Figure 6.4), although there are not the most popular. They have proven to be a good option for highly dynamic applications with large strokes and high positioning accuracy [63].

However, compared to other solutions, this system tends to be bulkier, heavier and the motor must be placed on the carriage in order to actuate the pinion. Moreover, they produce a high level of noise when compared to other drive systems which could repulse the patient. Thus, mostly due to space limitations the rack and pinion transmission system was excluded.

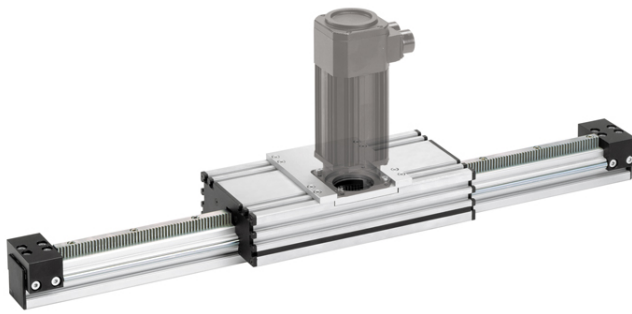


FIGURE 6.4: Linear unit with rack and pinion drive [63]

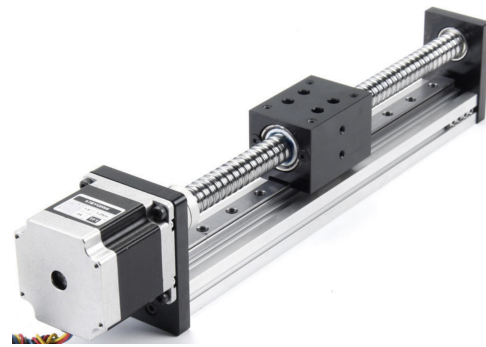


FIGURE 6.5: Ball screw linear actuator [64]

Ball screw assembly

This solution consists of a threaded shaft (screw) which provides a helical passage-way for a recirculating ball mechanism. When the screw is actuated in rotation, the ball assembly moves linearly.

Linear rails can be motorized thanks a ball screw system, by connecting a motor shaft to the screw and integrating the ball assembly in the carriage. Figure 6.5 shows a basic ball screw actuator, where the motor is placed on the left extremity longitudinally to the screw.

Ball screw linear actuators have a high efficiency ($\simeq 90\%$), requiring less motor torque than other screw systems [65]. They are generally the first choice for high position accuracy and big carriage loads.

In contrast, the stroke length is the major limitation of ball screw drive systems. As the length of the screw increases, the allowed speed decreases due to screw flexion. For the range of stroke that we were looking for, the existing solutions reaching the required velocity exceed the dimensional limitations previously established. Furthermore, ball screw drive technologies are considerably more expensive compared to other linear drive systems.

Toothed belt drive

In these systems, two pulleys are encircled by a tight flexible belt. The power is transmitted from the motor pulley to the second pulley via the belt thanks to its toothed surface.

By attaching a rail carriage to the upper fragment of the belt, linear motion is induced when one of the pulleys is activated. In Figure 6.6, we can see a belt-driven actuator from *Newmark Systems* company. For this kind of systems, the motor shaft is positioned perpendicularly to the guide rail.

Belt drive systems are probably best recognized for applications requiring long travel distances in a more cost-effective way, compared to ball screw actuators for a similar length.

Moreover, the belt-driven actuator provides very high efficiency (up to 93%) and practically inaudible due to the flexible nature of the belt [67]. Even if a re-tensioning



FIGURE 6.6: eBelt – Belt Driven Long Travel Linear Slide from *Newmark Systems* [66]

of the belt must be required time to time to ensure the proper transfer of torque, minor maintenance is needed for applications working with low loads [68] [69].

Transmission system selection

After proceeding this quick overview of the different linear motion solutions, we opted by a toothed belt drive system mainly for the following reasons:

- **Load capacity and position accuracy:** Working with low loads, our application can be perfectly handled by a belt drive system. Despite ball screw actuator allow a better position accuracy, it was not our major concern.
- **Compactness:** The desired maximal linear velocity could be provided by a ball screw actuator, but requires a much bulkier design compared to belt-driven models, for the same range of velocity.
- **Noise:** For medical purposes is really important to maintain the patient's trust and implication in the rehabilitation program. In fact, the noise caused by the system can be a frightening factor, demotivating the patient to work with such a device, especially in children and elderly people. In that sense, belt drive systems are preferable.
- **Price:** As developing the first prototype, a budget needs to be respected. Generally, belt drive systems are considerably cheaper than ball screw systems.

6.3 Selection of the belt-driven actuator

Since the market is crowded with a multitude of choices for toothed belt-driven actuators, we benefited from the advice of someone more specialized in the field, Julien Sapin, the Co-Founder and CTO *Axinesis*. In fact, he has been working on the development of a successful rehabilitation device equipped with a linear actuator, thus he possesses a clear perspective about the existing linear motion solutions and manufacturers. After presenting to him our requirements, he pointed us *BOSCH Rexroth* company since he had positive experiences with their technologies.

Another well-known linear units supplier for the UCLouvain's engineering department is *Igus*. Benoît Herman mentioned that they had already effectively collaborated with UCLouvain for several student engineering projects, offering attractive prices. So we decided to exploit solely the solutions proposed by these two companies. To fasten the process and get a price quote of the adequate actuator, we requested the technical help of an assistant from both companies.

After all, even if *BOSCH Rexroth* products are certainly more effective in accuracy, load capacity and durability, we considered their proposition too costly for the performances we were looking for. For nearly one-fifth of the price, *Igus* offers a satisfying and complete belt-driven transmission solution with the adapted stepper motor, the encoder, the controller and the required set of cabling (full proposition in Appendix C).

The selected technology, shown Figure 6.7, is the *drylin ZLW-1040-standard toothed belt axis* model, ideal for quiet and quick positioning of small loads. Moreover, with its main characteristics reported Table 6.8, is the second most compact system of the ZLW series. The guidance is ensured by a hard-anodized aluminum profile which supports the carriage thanks to self-lubricating sliding bearing in polymer [70]. The linear guide is actuated by a motor connected to the drive pulley through a spider coupling.

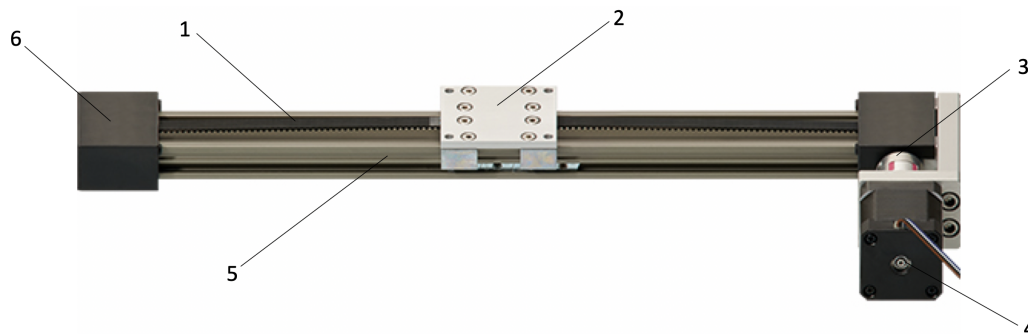


FIGURE 6.7: *Drylin ZLW-1040-standard toothed belt axis* model by *Igus* [71]

1. Polyurethane steel reinforced toothed belt
2. Carriage plate made in anodized aluminum
3. Spider coupling
4. Stepper motor
5. Hard-anodized aluminum profile
6. Pulley supports with deep groove bearings covered with a solid plastic housing

Total length	1004 mm
Width	74 mm
Height	45 mm
Carriage length	100 mm
Weight	2.12 kg
Max. radial stress	300 N
Max. speed	5 m/s
Max. position accuracy	± 0.2 mm

FIGURE 6.8: Linear actuator main specifications

6.4 Motor selection

Igus's catalog suggests a wide variety of motors compatible with this belt axis providing different levels of linear speeds and torques. Thus, selecting a motor from *Igus* was a straightforward choice. Very soon, their technical assistant communicated an engine for our application, thanks to a software which automatically identifies the suitable motor solutions according to the velocity and force requirements.

The chosen motor is a 2-phase hybrid stepper motor with an integrated encoder, operating at 24V, named NEMA23XL and illustrated Figure 6.9. It provides a maximal holding torque of 3.5 Nm and detent torque of 0.075 Nm.

To validate the motor choice, we could get an estimation of the maximal motor torque and velocity requirements, using the force and velocity data computed with *Robotran*. First, we derive an estimated value of the maximal motor torque as follow [72]:

$$T_{m,max} = \frac{F_{m,max} * R}{1000 * \varepsilon} = \frac{50 * 17}{1000 * 0.9} = 0.95 Nm$$

with, $F_{m,max}$: Maximal axial force in [N]

R : Radius of the drive pulley [mm]

ε : Efficiency of belt-driven actuator.

Naturally, the higher torques are required during the acceleration and deceleration phases of the trapezoidal velocity profile (section 5.3.1, Figure 5.4). In fact, the maximal linear speed is also reached (0.85 m/s) at the end of the acceleration profile. It corresponds to a maximal motor speed of around 480 RPM.



FIGURE 6.9: NEMA23XL stepper motor from *Igus* [73]

drylin® E stepper motor with connector, NEMA 23XL

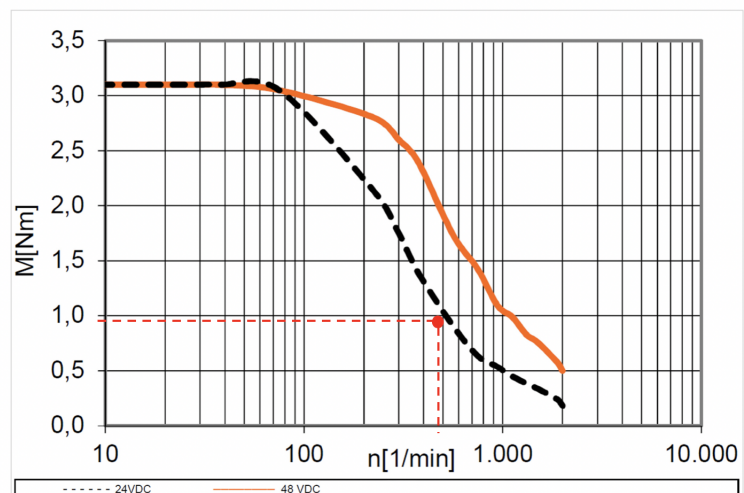


FIGURE 6.10: Speed - Torque curve of NEMA23XL in full capacity [74]

By taking a look at the Speed - Torque curve in full capacity (black dotted curve in Figure 6.10) of the selected stepper motor provided in the *Igus* catalog, the most

extreme operating point (480 RPN, 0.95 Nm) falls just under the curve. Taking into consideration that this motor is able to cover the most extreme point, kinetically and dynamically, it should be adequate to handle the needed range of torque and velocity in a normal scenario.

6.5 Linear guide deflection estimation

Early in the design, we should determine if the rail system could be carried by a single point of support in its extremity, by estimating the deflection generated by the vertical load. Otherwise, additional support must be provided to the rail.

Our linear actuator model has a permissible deflection up to 2 mm maximum to maintain an appropriate functioning [75]. This limit cannot be exceeded in any circumstances. Therefore, we will try to estimate the deformation of the linear guide, to see whether extra support is necessary.

The graph presented Figure 6.12, from the *Igus* catalog, gives the rail deflection according to the vertical load applied on the carriage and the length between the supports. It can be compared to a fixed beam clamped in both sides with a centered load as shown in Figure 6.11. However, in our case the linear guide will be supported in only one of its extremities (Figure 6.13), thus first, we need to estimate its Young Modulus.

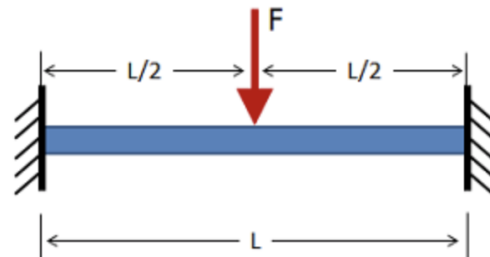


FIGURE 6.11: Fixed Beam with centered load [76]

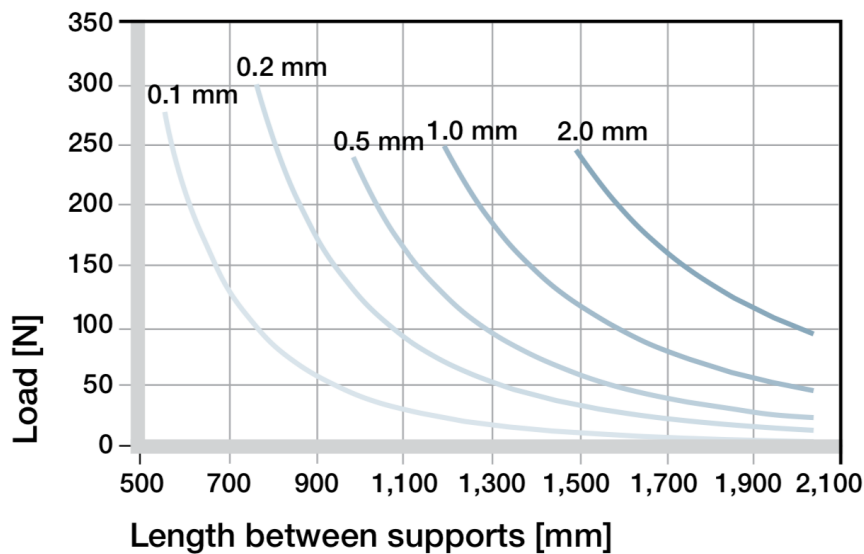


FIGURE 6.12: Deflection curves w.r.t the vertical load and the length between rail supports [75]

By considering one specific point in the graph (Load = 250N, Length = 1500 mm and deflection = 2 mm), the deflection equation for fixed beam with a centered load, can be written as follows [76]:

$$\delta = \frac{FL^3}{192EI} \Leftrightarrow E = \frac{FL^3}{192\delta I} = \frac{250 * 1.5^3}{192 * 0.002 * 2.7 * 10^{-8}} \simeq 81GPa$$

with, δ : Deflection in [m], in the middle of the rail

F : Vertical load in [N]

L : Length of the rail [m]

E : Young Modulus [GPa]

I : Second moment of area [m^4].

As mentioned before, in our design the rail is clamped in only one side, where the force F is the gravitational force of the leg and its support, and L is the stroke of the linear guide.

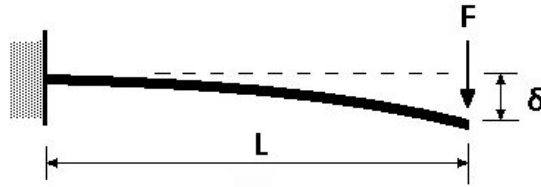


FIGURE 6.13: Cantilevered beam with punctual force [77]

For our configuration and dimensions, the maximal deflection is computed with the following formula [76] (Considering $F_{z,max} = 300N$ mentioned in section 6.1.3):

$$\begin{aligned} \delta_{max} &= \frac{F_{z,max}L^3}{3EI} = \frac{300 * 0.8^3}{3 * 81 * 10^9 * 2.7 * 10^{-8}} \\ &\simeq 0.0237m = 23.7mm \end{aligned}$$

In fact, the resulted deflection is way above the permissible deflection, so additional support can not be avoided.

Chapter 7

Instrumentation components

This chapter presents the selection and function of the measuring tools required to control the system. Three types of sensors are described, namely, angular encoders, force transducers, and contact switches.

7.1 Abduction angular position

The first type of data that the device has to identify is regarding the patient positioning. When a patient's leg is mobilized, the device must determine the hip angular position (θ_{hip}) at any time during the abduction/adduction movement. Indeed, this information will be fundamental to the controlling system. Before the selection of the measuring technologies, we first identify what parameters should be measured to estimate the leg angular aperture.

7.1.1 Geometrical assessment

Figure 7.1 is a simplified geometrical drawing of the device from above, at its initial configuration and at a random running configuration.

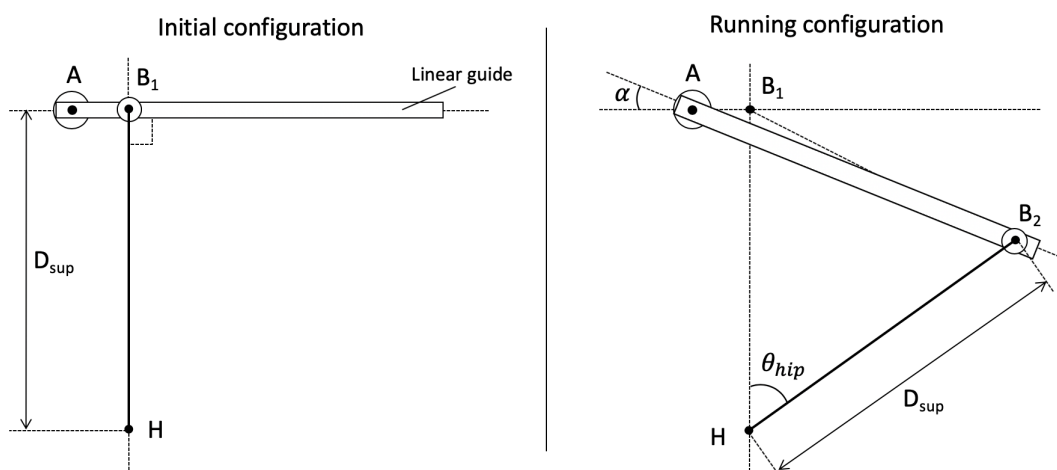


FIGURE 7.1: Top view of a geometrical drawing of the actuation system

- H : Hip joint position
- D_{sup} : Distance between the hip joint and the Joint B.
- A : Joint A position
- B_1 : Joint B initial position
- B_2 : Joint B running position
- θ_{hip} : Hip abduction angle
- α : Angle travelled by Joint A

In the initial configuration, we assume that the linear guide and the longitudinal axis of the leg (HB_1) are perpendicular to each other. In addition, the distance D_{sup} is always constant throughout the abduction cycle, which forms an isosceles triangle HB_1B_2 . Moreover, the pivot A is fixed, while the joint B travels from B_1 to B_2 along the linear guide.

To better visualize the problem, an additional drawing in the running configuration is provided Figure 7.2. Point C is the intersection between AB_2 and HB_1 .

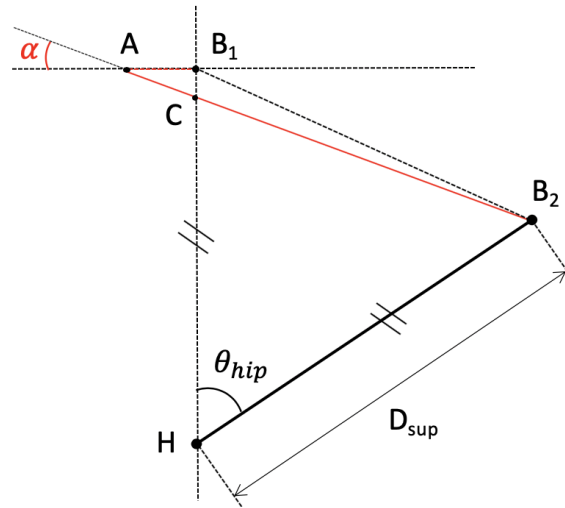


FIGURE 7.2: Geometrical drawing in running configuration

Based on this drawing, one way to mathematical express θ_{hip} and D_{sup} according to other parameters, is applying triangle basic rules, such as the laws of sines and cosines [78]. That way, within the different triangles, the following relationships were established:

Triangle (AB_1B_2)

- Cosines law:

$$\begin{aligned}
 (B_1B_2)^2 &= (AB_1)^2 + (AB_2)^2 - 2 * AB_1 * AB_2 * \cos(\alpha) \\
 &\iff \\
 B_1B_2 &= \sqrt{(AB_1)^2 + (AB_2)^2 - 2 * AB_1 * AB_2 * \cos(\alpha)}
 \end{aligned} \tag{7.1}$$

Triangle (CB_1B_2), given that:

$$\begin{aligned}\angle B_1AB_2 &= 90 + \alpha \\ CB_2 &= AB_2 - \frac{AB_1}{\cos(\alpha)} \\ \sin(90 + \alpha) &= \cos(\alpha)\end{aligned}$$

- Sines law:

$$\begin{aligned}\frac{CB_2}{\sin(\angle CB_1B_2)} &= \frac{B_1B_2}{\sin(\angle B_1CB_2)} \\ &\iff \\ \angle CB_1B_2 &= \arcsin\left(\frac{\cos(\alpha) * (AB_2 - \frac{AB_1}{\cos(\alpha)})}{B_1B_2}\right) \\ \angle CB_1B_2 &= \arcsin\left(\frac{\cos(\alpha) * AB_2 - AB_1}{B_1B_2}\right)\end{aligned}\tag{7.2}$$

Triangle isosceles (HB_1B_2)

- Sum of triangle angles = 180° :

$$\theta_{hip} = 180 - 2 * \angle AB_1B_2\tag{7.3}$$

- Cosines law:

$$\begin{aligned}(B_1B_2)^2 &= 2 * D_{sup}^2 - D_{sup}^2 * \cos(\theta_{hip}) \\ &\iff \\ D_{sup} &= \sqrt{\frac{(B_1B_2)^2}{2(1 - \cos(\theta_{hip}))}}\end{aligned}\tag{7.4}$$

By replacing equation 7.1 in equation 7.2, and then inserting the resulted equation 7.2 in equation 7.3, the final expression of θ_{hip} is:

$$\theta_{hip} = 180 - 2 * \arcsin\left(\frac{\cos(\alpha) * AB_2 - AB_1}{\sqrt{(AB_1)^2 + (AB_2)^2 - 2 * AB_1 * AB_2 * \cos(\alpha)}}\right)\tag{7.5}$$

Thus, θ_{hip} depends on: AB_1 , AB_2 and α (shown in red Figure 7.2).

First, AB_1 corresponds to the initial distance between both pivots, which is intrinsic to the device design and thus known for every patient morphology.

Then, AB_2 is the distance covered by the carriage added to AB_1 . In practice, its value is obtained thanks to the incremental encoder coupled to the motor (detailed in the next section 7.1.2). Knowing the angular position of the motor pulley and its radius, allow to easily deduce the distance traveled linearly by the carriage in the rail system at any moment.

Finally, one angular measurement is needed. In fact, this parameter can be measured from one of the two revolute joints. We selected α as the measured angle, since the integration of an angular encoder in the Joint A will cause less wiring complications than in Joint B.

For instance, if we consider that $AB_1 = 100$ mm and that the carriage travelled a distance AB_2 of 600 mm, while α is 20° , it gives a hip abduction angle of 46.5° and $D_{sup} = 592$ mm.

7.1.2 Incremental encoders

The encoders integrated into both the *NEMA23XL* motor and in one of the passive rotational joint, are angular incremental encoders following the same working principle.

Inside such encoder, there is a disc containing a specific pattern of opaque and transparent bands. When rotating, thanks to a sequence of on and off signals that are induced by the optical disc, the movement from one position to another can be detected. However, contrarily to absolute encoders, the angular position is lost if power is cut [79].

Three signals are measured A, B, and N, which consist of pulse squared signals, shown in Figure 7.3. As channel A and B are shifted, their order of detection determines the direction of rotation. In addition, the N channel induces only one pulse per revolution, indicating the zero position. Moreover, the inverted copies A/, B/ and N/ are used to provide better noise robustness at the receiver [74]. Finally, the encoder's resolution is measured in pulses per revolution (PPR).

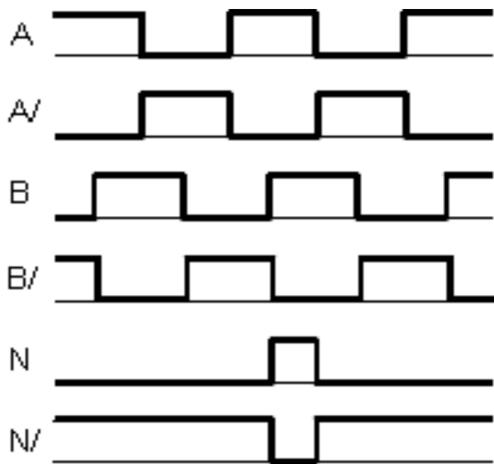


FIGURE 7.3: Output quadrature signals from an incremental encoder [74]



FIGURE 7.4: Incremental encoder AM102-V from CUI [80]

Regarding the motor's encoder, based on 500 PPR, 2000 states per revolution are obtained, thanks to the combinations of A and B quadrature signals. It gives an angular accuracy of $0.18^\circ/\text{state}$. Considering a pulley diameter of 30 mm, it leads to a carriage position accuracy of 2.7 mm/state. However, depending on the carried loads,

its actual position and the measured position will slightly differ due to the flexible nature of the belt. Thus, the overall accuracy of the computed distance may be affected accordingly.

For the measurement of the α angle, an incremental encoder from the series *AMT10* by *CUI* was an effective choice. It provides a very slim design (Figure 7.4) and is compatible with a wide range of shaft sizes, for an attractive price. In addition, the selected model has a resolution of 2048 PPR, which leads to an angular accuracy lower than $0.05^\circ/\text{state}$ [80].

7.2 Force measurement

As mentioned in the technical requirements, M-VITEC needs to measure the force generated by the patient's leg. In fact, unlike conventional CPM devices, our device works in an active mode where the driving command is adapted to following a pre-defined trajectory according to the patient efforts. For our purposes, among the three components of the force, two are interesting to acquire: the normal force and the axial force with respect to the leg.

As the angle between the linear actuator and the leg varies along the abduction cycle, the driving force will not be fully effective, transferring a normal and axial force to the leg.

Abduction around the hip is produced by the component pushing always normal to the leg. Thus, by accessing the interaction force in the normal direction, the device can adjust the motor command to achieve the most constant hip angular velocity profile. In addition, it would provide an estimation of the force developed by the abductor muscles and monitor patient muscular strength.

Along with the normal force, measuring the axial force could be useful for safety reasons. Having a real-time measure of both forces enables the device to be reactive to abnormal forces, by shutting down the actuator when they go above a given threshold.

Finally, these measures would be valuable to validate the model handled with *Robo-tran*, by comparing the simulation results with the real measured forces.

For this prototype, only the normal force will be acquired, essential to control the machine. In practice, uni-axial force sensors are more compact, easy to purchase (without special request) and thus considerably cheaper compared to bi-axial sensors. Nevertheless, implementing a bi-axial force sensor could be an asset for further M-VITEC versions.

The chosen sensor is a miniature tension/compression force transducer from *TEC-SIS*, shown figure 7.6 [82]. Inside, it contains a strain gauge which is an electrical conductor with a resistance sensible to small deformations. As illustrated in Figure 7.5, when the gauge elongates, resistance increase and, inversely, when the gauge compresses, resistance reduces.

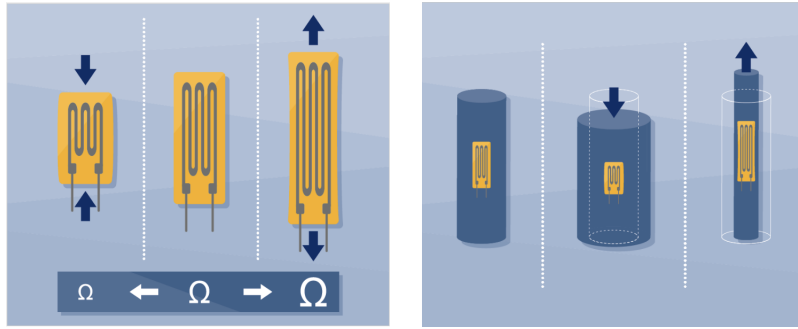


FIGURE 7.5: Strain gauge and spring element compression/extension and the influence in electrical resistance [81]

The gauge is firmly attached to a spring element (usually a metal piece) which deform together. By measuring an output voltage, the corresponding electrical resistance is obtained. This latter is proportional to the strain on the material, which is also proportional to mechanical stresses. Based on the properties of the spring component the applied force can be computed [83] [81].

As we look at Figure 7.6, the selected model is composed of a cylindrical body which senses the compression and extension forces applied on the two threaded bolts. It offers a wide range of force capacity, from 1.5N to 5000N. For our concerns, a rated force F_{nom} of 200N is surely sufficient given the normal force amplitudes. In fact, this sensor has a limit force of $150\%F_{nom}$, leading to a reliable measuring until a 300N. Additionally, the breaking force for the transducer is 600N, letting us a good safety margin. Finally, it provides measurement with a relative linearity error of $\pm 0.5\%F_{nom}$, which corresponds to an error of less than one newton throughout the complete force scale [84].



FIGURE 7.6: Miniature tension/compression force transducer from TECSIS[84]

7.3 Contact switches

7.3.1 Calibration switches

This switch aims to calibrate the angular position of the linear guide at each abduction cycle and thus, improve the angular measurements. Ideally, it must inform the control system when the linear guide is at the 0° position.

Concerning proximity sensors, there is a multitude of solutions on the market that require physical contact or operate without contact. We opt by snap-action switch mainly due to their low cost, their high durability, and the availability in very small

sizes [85].

The used model comes from *Marquardt Switches* company and consist of an open electrical circuit enclosed in a very small plastic housing with a push-button, as can be seen in Figure 7.7 [86].



FIGURE 7.7: "Single Pole Single Throw" snap-action switch from *Marquardt Switches* [86]



FIGURE 7.8: "Single Pole Double Throw" snap-action switch with roller arm from *Wurth Elektronik*[87]

The circuit is normally open (NO), thus no current flows when the button is in resting position. Once an external force is applied on the button, the circuit closes and a signal can be measured [88].

7.3.2 Stroke limit switches

Moreover, for safety reasons a proximity sensor was integrated into the linear guide system. This sensor is intended to be triggered when the carriage is close to both ends of the planned stroke. Thereafter, inside of theses end stroke zones, a slowdown command (for instance limited velocity) can be imposed to the carriage, to make sure dangerous situations are avoided.

The selected micro-switch from *Wurth Elektronik* is similar to the previous but with a different actuation mechanism. As illustrated Figure 7.8, it contains a lever arm with a roller, which sits over the push-button. This mechanism allowed to integrate only one switch over the carriage instead of two switches in each end of the linear guide (more details in section 8.7).

Chapter 8

Detailed design

This chapter describes in detail the design and assembly of the components composing the final prototype.

First of all, a general overview of the device and its main parts are presented with the general dimensions. Further, we will zoom in on each sub-assembly and explain how the joints and sensors were integrated.

While selecting and assembling the components, we always tried to reach the most compact design, mainly in height. Most of the parts come from *MISUMI* catalog which contains a wide variety of products, at low prices. Hence, most of the components are geometrically configurable, leading to an easy assembly.

Throughout this chapter, the numbers in brackets included in the text, refer to the components shown in the figures. More details (supplier, ref., price, etc.) of each piece are provided in Appendix D.

All the 3D representations were made using the CAD software *SolidWorks*, by *Dassault Systèmes* [53].

8.1 General presentation of M-Vitec

A representation of the final prototype in the initial configuration, installed on a massage table (similar dimensions than physiotherapy table), is given in Figure 8.1. The device should be placed, along the y-axis, such that the patient's heel touches the bottom of the leg support.

Then, M-Vitec must occupy roughly half of the table's width (x-axis), leaving sufficient room laterally for the healthy leg. Consequently, for a standard physiotherapy table, a large part of the base ($\simeq 3/4$ of the total width) falls outside. Thus, extra support is provided by a telescopic leg, anchored in the brimming extremity of the base and standing on the room's floor. This leg can be adjusted according to the table height, as well as folded and blocked for easier transportation of the device. The extended design of its foot provides stability to the device.

Figure 8.2 shows a top and front view of the device, as well as its main dimensions. M-Vitec can be divided into three principal parts: the base, the motorized linear guide, and the leg support.

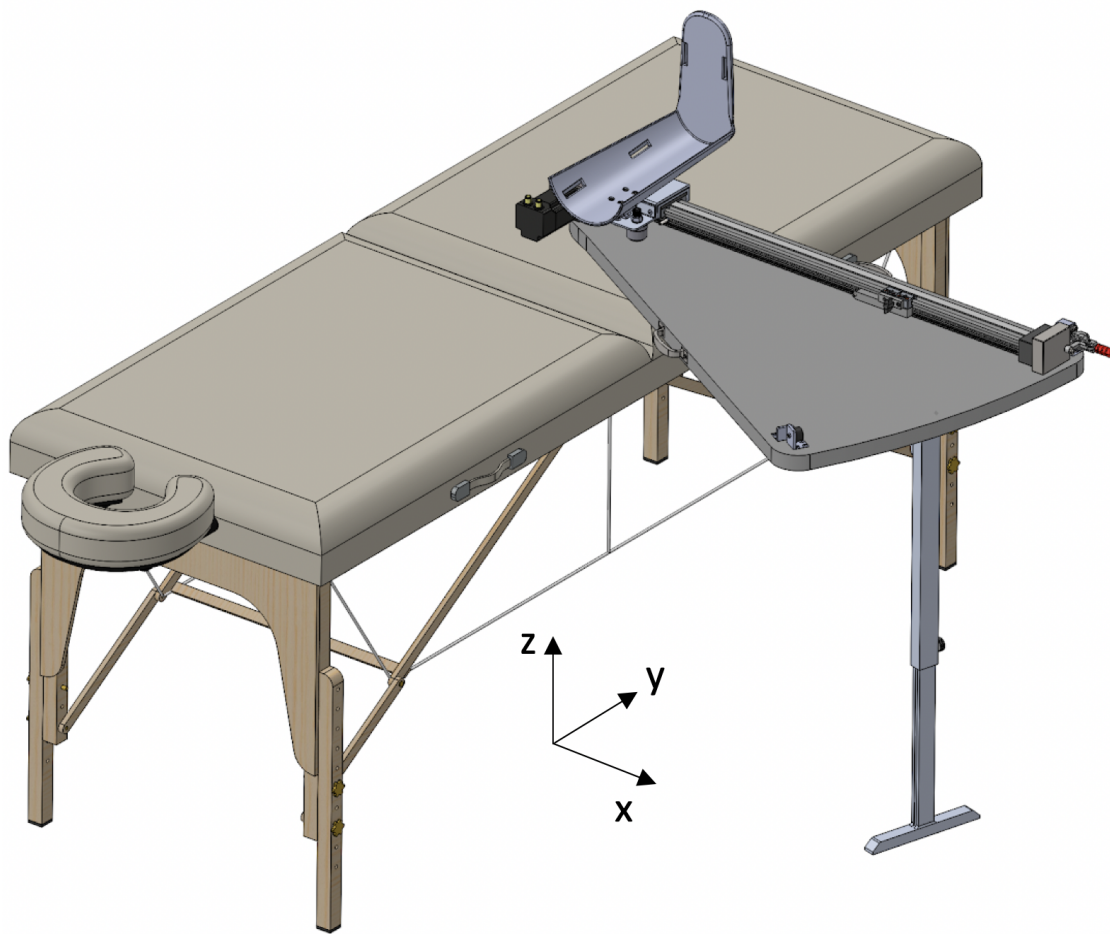


FIGURE 8.1: Isometric view of M-Vitec installed on a massage table [89]

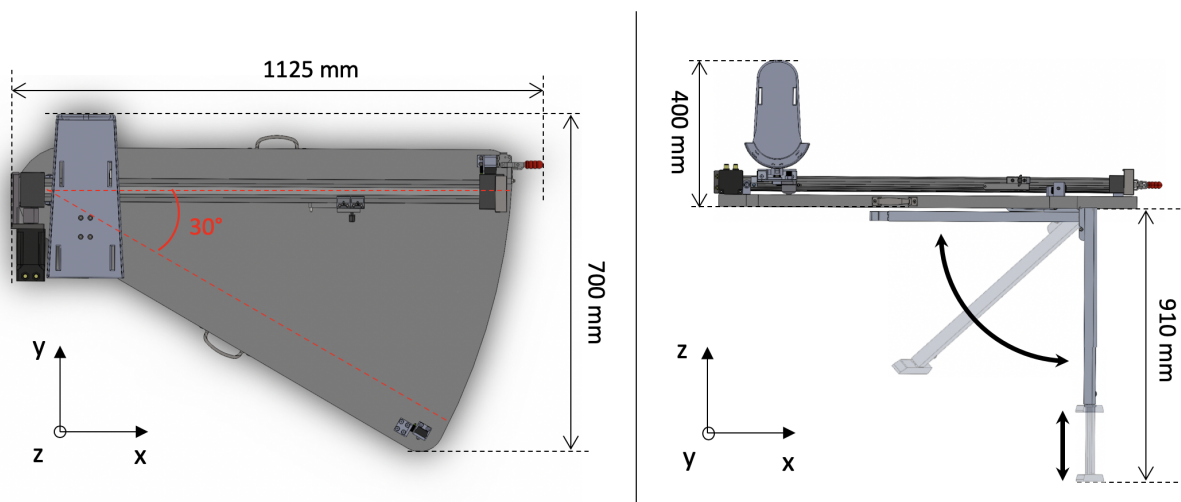


FIGURE 8.2: Top view and front view of M-Vitec along with its main dimensions

Firstly, the base consists of a flat plastic (polyamide) sheet which adopts a circular arc shape with 30° . This special form was imposed by the angular stroke of the linear guide, which provides a sliding surface for the rail rolling support (see section 8.6). That way, compared to a standard rectangular sheet, the bulkiness and weight were minimized. However, it leads to a more complex manufacture.

Two metallic handles were added on both sides of the base to facilitate the positioning of the device under the patient and the transport.

Secondly, one end of the belt-driven linear guide is connected to the base through two revolute joints. Thus, the rail can rotate around both the z and y-axis. Besides, the motor and its coupling were placed near the pivot, otherwise, it would induce important inertial forces and weight in the rotary guide. The actuated carriage travels over the rail and pushes linearly the leg support.

Finally, to allow the leg support to always stay aligned with the patient's leg, the last revolute joint was implemented next to the carriage. In addition, a rolling component was integrated under the leg support to handle the weight applied to it during all the abduction cycle.

The prototype is 1.125 m long, 0.7 m wide and 0.4 m high (with the folding extensible leg retracted). In addition, when the machine is positioned under the patient, its lower limb is elevated by only 10.4 cm which guarantees a low hip flexion (not recommended shortly after THA). Furthermore, the device weighs approximately 18.7 kg.

According to the prices of the components used in this prototype, the total purchase cost is 1368 €. Nonetheless, the price of some parts was unknown (for instance the force sensor, the base, and the leg support) and thus not considered in this amount.

The following sections described in detail the assembly of most of the components.

8.2 "Base-Guide" joints

Two rotational degrees of freedom are permitted to the linear guide.

Guide z-axis joint

This joint is placed under one extremity of the linear guide and ensures its rotation with respect to the base, around the z-axis. Among a multitude of possibilities to generate a rotational motion, we chose a simple bearing unit coupled with a compatible shaft, both from *MISUMI* [90] [91].

The unit contains a double-shielded deep groove ball bearing allowing the rotational motion between the shaft (6) and the housing (4). Its flanged design facilitates the attachment to the device's base (5) with screws (7), inside a circular slot.

Regarding load capacity, this type of bearings tolerates high radial loads and limited axial loads. Radially, the loads applied to the bearings will be similar to the forces

generated by the motor, by the action-reaction effect, thus not exceeding much more than 50 N. Axially, since most of the vertical load is carried by the rolling support (described section 8.6), these bearings will only experience a small load caused by the weight of the motor, as well as a small portion of the rail's weight. Thus, the efforts applied on the bearings are comfortably bearable.

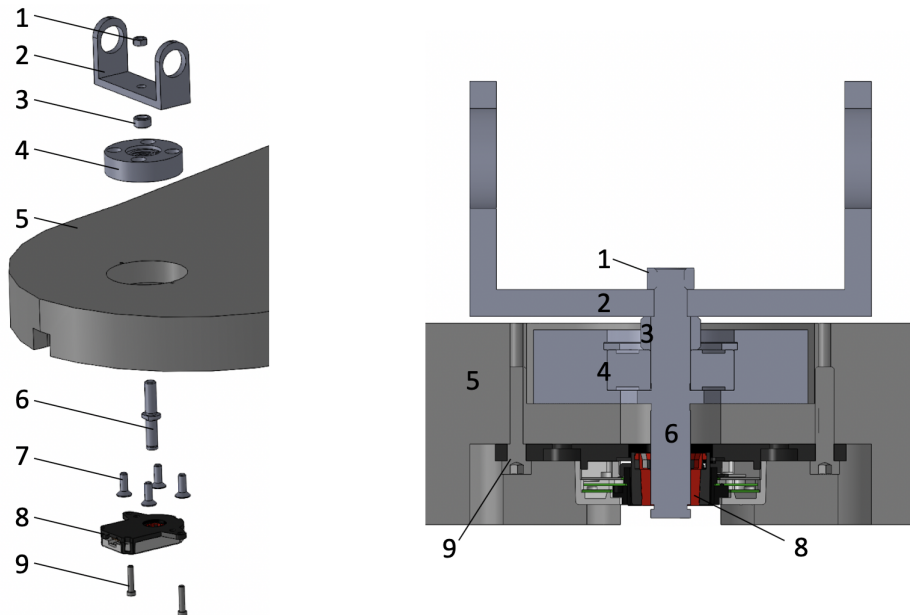


FIGURE 8.3: Exploded and cross-section views of the z-axis joint assembly

The shaft is a short cantilever steel piece (6), with one threaded end and a retaining ring groove at the other end. As we can see in Figure 8.3, it contains a shoulder which axially blocks the shaft and the bearing, in the upward direction. The threaded end serves to attach the shaft to a U-shaped aluminum piece (2) (supports and fixes the linear guide), as the other end is inserted in the bearing unit. A bearing spacer (3) enables to effectively level-up the U-shaped part (2) so that the linear guide does not touch the device base (5). To strengthen the assembly, a compact nut (1) is fastened from above.

Finally, the AMT10 incremental encoder (8) is fixed under the base of the device with two screws (9). The lower shaft end is inserted inside the encoder opening, thus its angular position is measured.

Guide y-axis joint

This joint also enables an angular movement of the linear guide, but this time around the y-axis.

In reality, it improves the mechanical robustness of the system. In the absence of such degree of freedom, the guide needs to be perfectly aligned and heightened such that the rolling support sits correctly on the sliding surface. A slight misalignment could put the linear guide in a cantilevered configuration (rolling support not acting anymore) generating a considerable torque within the ball bearing unit. Since a single

ball bearing unit is not designed to handle torques, the mechanism could potentially fail. Thus, as a precaution, this rotary joint was integrated.

Looking at Figure 8.4, the rotational degree of freedom is permitted between the rail (14) and the U-shaped component (2). Two deep groove bearings (11) were placed in each arm of the U-shaped piece (2). Moreover, slider nuts (13) are inserted into the side grooves of the aluminum rail profile (14). Then, two pin pivots with a threaded end (10) go through the bearings (11) and are screwed, from both sides, to the sliders (13). A bearing spacer (12) is used in each of the sides to separate the rail surface (14) and the bearings (11).

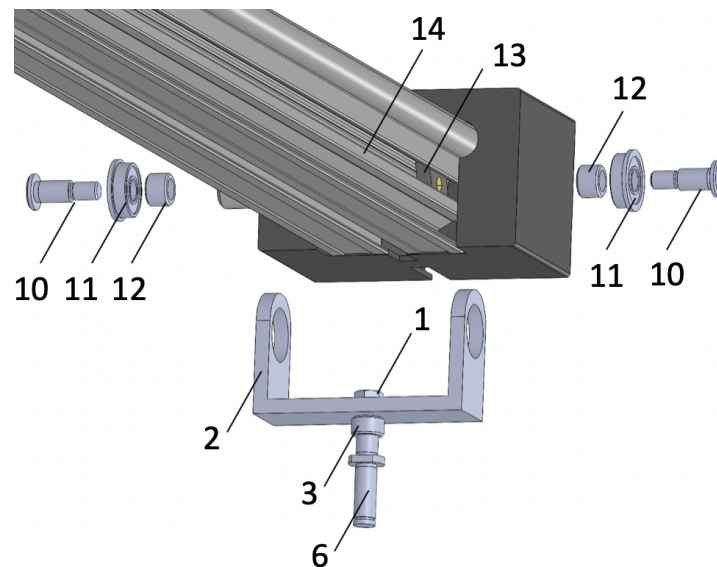


FIGURE 8.4: Exploded view of the y-axis joint assembly

8.3 "Carriage-Leg support" detachable connection

Over the rail system, the driven carriage (15) travels linearly moving the lower limb and its fixation. As one of the main objectives is to lower the height of the device, the leg support and the remaining pivot were placed on the side of the rail instead of stacking over it.

Therefore, as represented in Figure 8.5, a short metal channel (17) is unified to the carriage (15), providing an attachment for an intermediary piece (18) that expands to the side of the rail (14). Then, a knurled knob (19) is used to attach and detach the side rail platform (18 and 20) and the carriage. This allows to easily separate the leg support (not represented here) from the base of the device for transportation, as well as when switching from one leg to the other. In fact, as the leg support pivot is no longer aligned with the rail, the orientation of the leg support must be adapted according to the body side that is being treated.

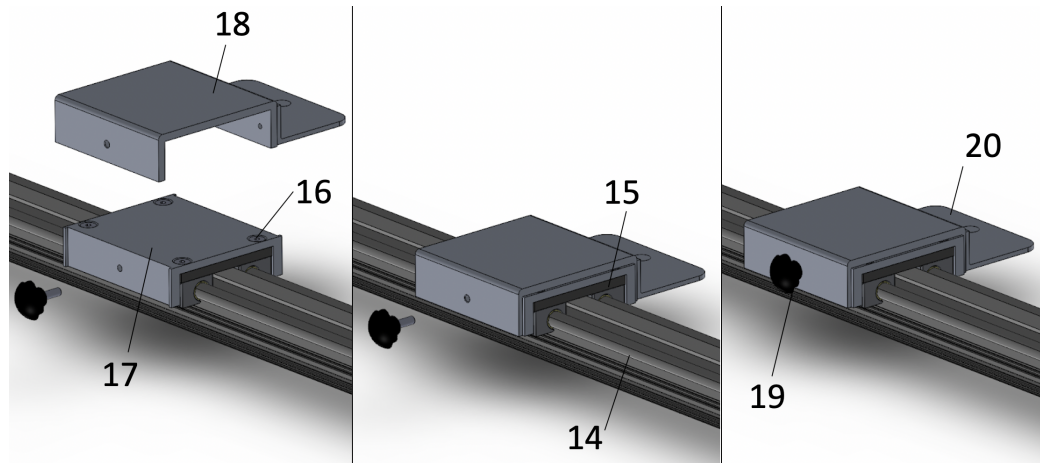


FIGURE 8.5: Removable system with knurled knob within the carriage

8.4 "Carriage-Leg support" revolute joint

This rotary joint sits over the lateral platform (20), ensuring the rotation of the leg support with respect to the linear guide (14). Again, we were looking for a flattened design which handles a maximal vertical load of 300 N. Other applied loads and torques were sufficiently small to be considered.

As for the rail yaw joint, a single deep ball bearing unit was an effective solution, reaching low friction levels in a very compact way. The bearing unit (22) was screwed on the downside of the platform (20) leaving space above for the remaining components.

Furthermore, a pin pivot with a threaded end (24) is inserted through a metallic band (29) and the bearing (22). A nut (27) blocks vertically the whole. A spacer (26) separates the bearing (22) from the band (29) allowing them to rotate, one over the other (Figure 8.7, Step 1 and 2).

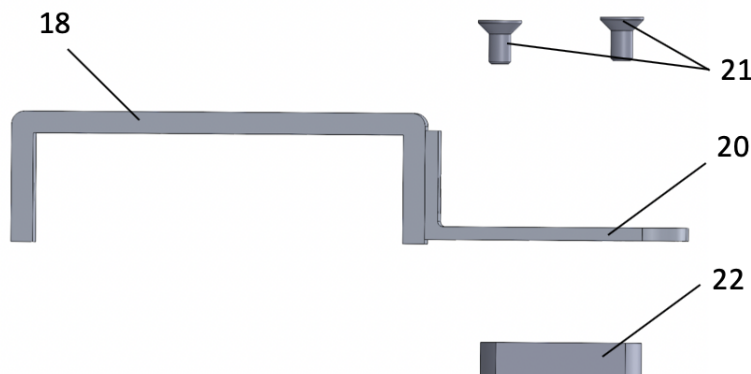


FIGURE 8.6: Side view of the carriage lateral platform and bearing unit assembly

8.5 Force sensor mounting

The selected force sensor only acquires the compression and traction forces in one direction and is not designed to bear efforts in other directions. Otherwise, it would cause unintended deformations of the strain gauge and disrupt the measurements. Additionally, high lateral loads would induce shear stresses that could break the sensor.

Thus, the sensor was integrated such as, it carries the totality of the normal load but does not sense efforts in other directions.

For that purpose, a miniature ball linear guide (28) was installed on the top of the revolute joint. Then, as illustrated in Figure 8.7 (Step 3 and 4), a rigid U-shaped piece (32) is bound with a small carriage (31), linearly slides on the guide.

The sensor (33) is mounted along the axis of the guide, as shown Figure in 8.8. In fact, one of its threaded rods is linked to a branch of the U-shaped part (32), while the other is clamped to a small steel piece (34). That way, the sensor blocks the linear movements of the carriage (31) and thus, can measure the normal forces. In addition, as the ball guide system (28 and 31) causes insignificant friction in the sliding direction, the measurements are not perturbed while handling vertical and axial efforts.

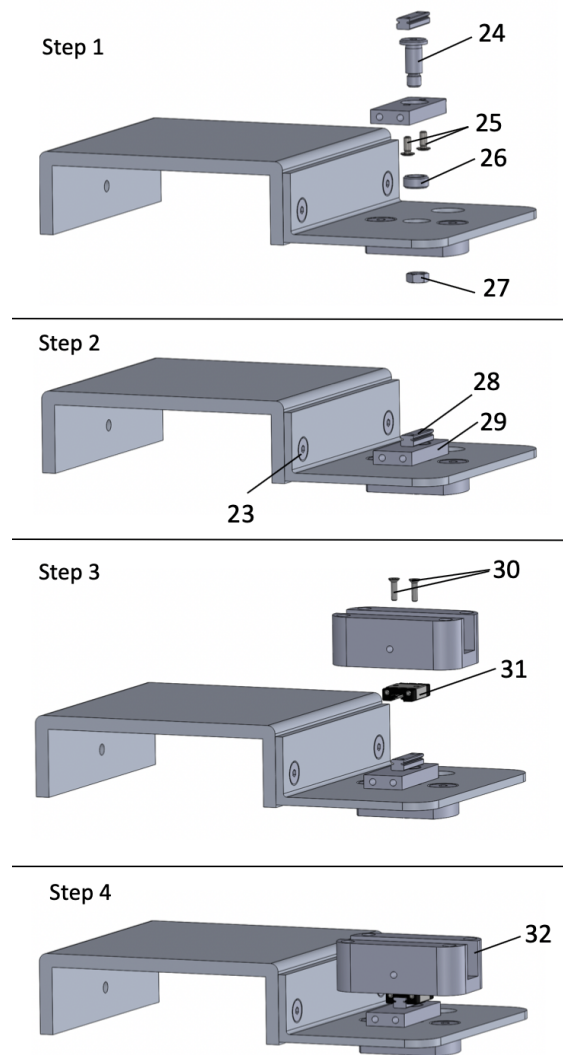


FIGURE 8.7: Pivot pin and bearing unit assembly (Step 1 and 2). Linear guide system assembly for the force sensor mounting (Step 3 and 4)

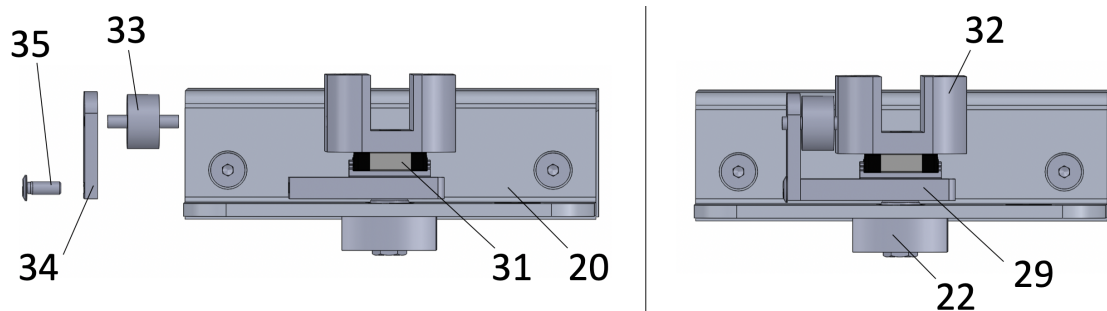


FIGURE 8.8: Side view of the force sensor (33) mounting

8.6 Rolling support

As demonstrated in section 6.5, the rail cannot withstand all the vertical load, a convenient solution was to install a support component connected to the carriage. Therefore, as the carriage moves back and forth, the leg weight is always taken over by this support, which decreases considerably the moment generated in the clamped end of the rail.

To follow the carriage movements, this support must roll over a sliding surface. Therefore, we used a multi-directional support component, called ball transfer unit (38), that moves over the smooth surface provided by the base.

Such an element is composed of a load-bearing ball supported by a layer of small steel balls encapsulated in a housing. These rolling components drastically reduce friction and allow heavy loads to be moved with a minimum of effort [92].

The chosen model IK-19SNM (38), are supplied by *IGUCHI KIKO* company and can support until 588 N. This caster is especially designed to ball-down position applications and contains a threaded rod, which facilitates its mounting [93].

The unit was fixed to the side platform (20), next to the revolute joint with a classic M10 nut (36). Its threaded end was covered with a rubber bumper to amortize the possible collision with the rotatory U-shaped piece (32) (Figure 8.9).

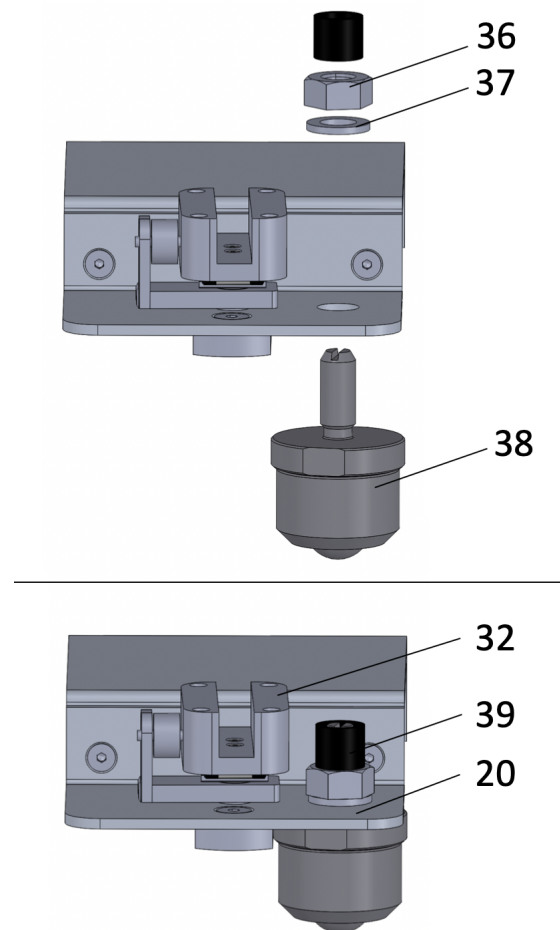


FIGURE 8.9: Ball transfer unit mounted in the rail carriage

8.7 Linear guide blocks & micro-switches

In Figure 8.10, we can see a segment of the linear guide, corresponding to the beginning of the carriage stroke. Under the carriage, the micro-switch with a roller lever arm (46) was screwed to the rail side platform (20). Moreover, two threaded rods with a soft tip (42) are fixed to the carriage side cavities and act as the switch actuators. In fact, any time the carriage crosses these two screws, the tip pushes the roller which activates the switch. Hence, the system is informed that the carriage is approaching the stroke limit. Closer to the motor, a mechanical stop (49) is attached to the rail, blocking the carriage longitudinally.

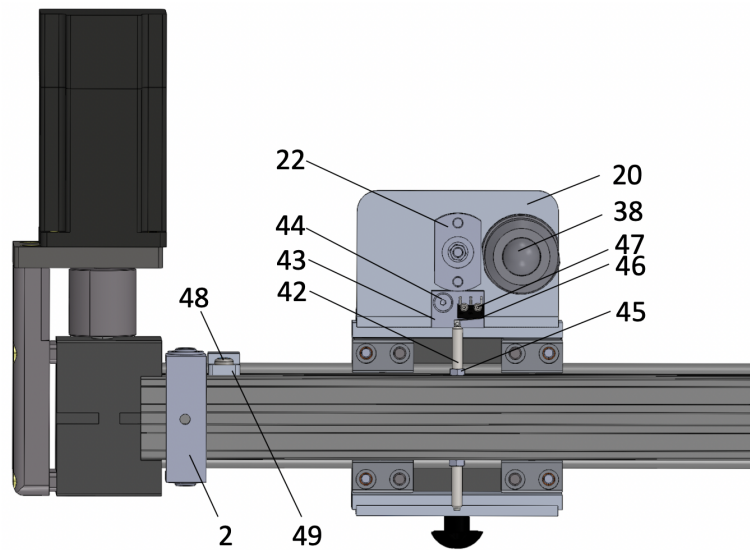


FIGURE 8.10: Beginning of the linear guide: Block and micro-switch system mounting

Additionally, for safety reasons, a pillow block (51) was integrated to the linear guide (14) to limit the carriage stroke and thus define the maximum angle of abduction (Figure 8.11). This component (51) from *Igus*, can be manually adjusted along the rail, thanks to a tightening knob [94]. It should be positioned by the physiotherapist after evaluating the patient's ROM.

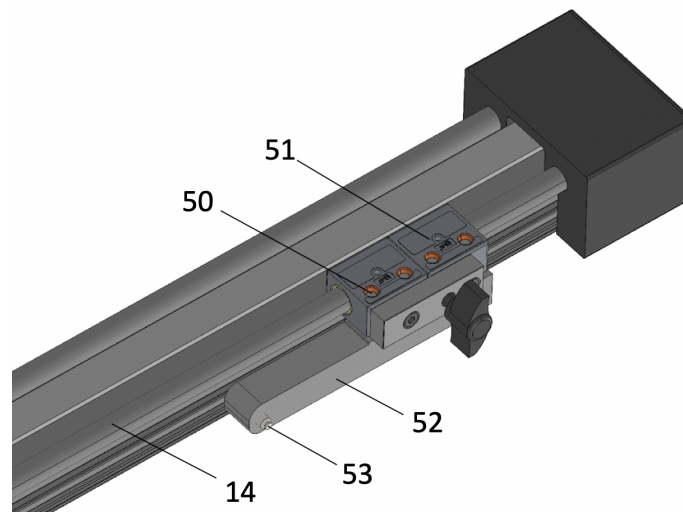


FIGURE 8.11: End of the linear guide: Adjustable pillow block

Again, a threaded rod with a soft tip (53) is linked to the stroke block (51) via an extended bar (52). By touching the micro-switch mounted on the carriage, it alerts the system that the maximum stroke will be shortly reached. Hence, an emergency stop is triggered to avoid a collision between the carriage and the block (51).

8.8 Leg support

The man-machine interaction takes place in the leg support. This latter mobilizes the lower limb extremity (calf and foot) during all the abduction exercise.

The final design, presented Figure 8.12, consists of a plastic shell (39), slightly conical, mimicking the calf morphology. The foot sits on a perpendicular surface which always maintains it in an upright position. The width and depth of the shell were fixed making sure every leg shape suits inside. Moreover, three pairs of slots serve as an attachment for adjustable Velcro straps which fasten all types of legs in an easy and fast way. In addition, the sharp edges were rounded to avoid any risk of harming the patient during the exercises.

Such support could be made of a thermoplastic polymer (for instance, PET) providing stiff and strong mechanical properties, that can be manufactured with a 3D printer. For a smoother contact with the patient, the inside face of the shell could be covered with a foam coating.

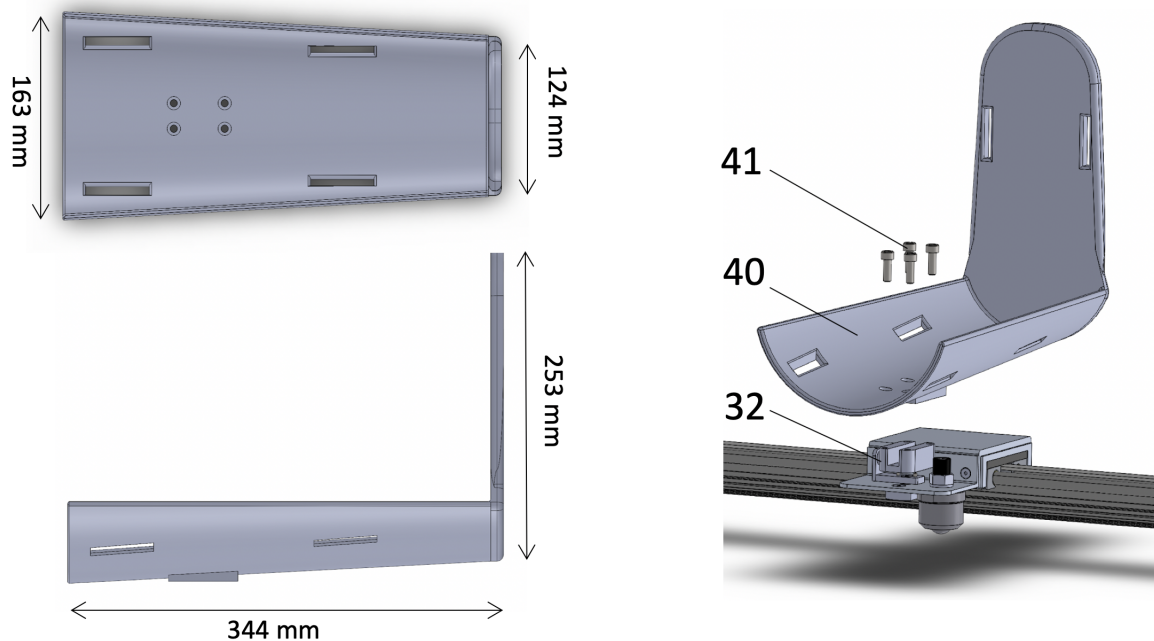


FIGURE 8.12: Leg support design and assembly

The leg support (39) is mounted perpendicularly to the force sensor over the U-shaped piece (32) thanks to four screws (40), shown figure 8.12. Hence, the force sensor is always normal to the longitudinal axis of the leg, allowing to collect the normal forces generated by the patient.

8.9 Rail locking system & angular switches

To avoid the guide to freely rotate when the device is not in use, and mostly for safe transportation, a locking system was implemented. As illustrated Figure 8.13, it consists of a vertical hold-down toggle clamp (56), fixed parallel to the base (5). The clamp's arm transports a wedge-shaped part (54) which encapsulates the end of the linear guide, blocking it vertically and horizontally.

Moreover, shock-absorbing bumpers (62), limit the angular movement of the guide during the abduction cycles. Next to it, calibration switches from *Marquardt Switches* (59) were precisely positioned so that they are stimulated when the guide reaches its angular limit. That way, any time they are pushed, the system can reset the angular measurements, canceling the possible accumulated error.

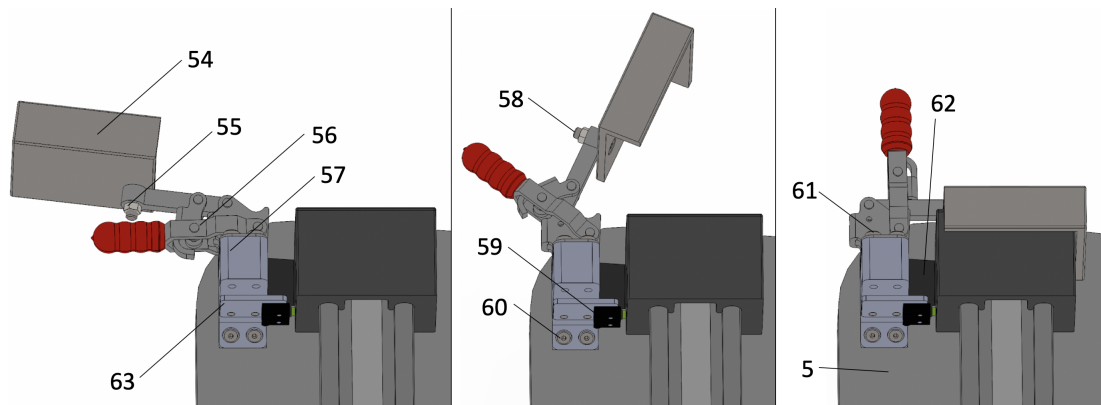


FIGURE 8.13: Guide locking system and calibration switch

Chapter 9

Conclusion

This conclusion highlights the main steps and achievements resulted from the past few months of work and suggests the key aspects requiring further investigation, to reach the most complete and efficient therapeutic device.

9.1 Main steps & achievements

The progression of this thesis leading to the final prototype can be divided into three parts.

The first part, regarding the state of the art of the hip rehabilitation practices, allowed to identify the root of the clinical need, while understanding the gaps of the current robotics devices aiming to treat hip muscular insufficiency.

In fact, the implantation of a prosthetic hip has proven to be a successful solution for a patient with a severely damaged joint. Despite the fact that the overall quality of life is improved, the muscular capacities around the operated hip cannot match the ones of their untouched hip. Naturally, this deficiency disturbs the stabilization function of these muscles, often leading to a long term limping.

Hence, the efficiency of the conventional therapies has been questioned and new trends are arising, such as therapeutic rehabilitation devices. However, we noticed that existing solutions in the market are limited to passive movements, not adjusting the trajectory according to a patient's behavior. Several projects have been trying to address this limitation, but their prototypes are difficult to transport and do not minimize the stresses inside the hip.

The conceptual design phase constitutes the second part of this work, where a large variety of solutions were considered in a detailed classification process.

The LMECA2855 course results were exploited and served as a basis to generate an exhaustive set of planar solutions. The *Robotran* software was a crucial tool to quantify the efforts generated by each of the mechanisms in the hip joint and the required driven force. By integrating the results of this analysis with several criteria, regarding the price, bulkiness and safety, a final actuation solution was achieved.

The last part concerns the dimensioning, selection and assembly of the mechanical and electronic components.

After identifying the technical requirements of the linear actuation system and navigating through an overview of the existing linear transmission system, we concluded

that a toothed-belt driven system is the most adapted solution for this application. In fact, it leads to a more compact and quiet alternative, providing high speeds, for an interesting price.

Regarding the instrumentation of the device, the geometrical assessment confirmed that hip abduction angle can be derived thanks to a combination of two angular encoders. Furthermore, measuring the force applied by the patient is what distinguishes M-VITEC from existing stationary rehab machines. Indeed, the device will be able to adjust, in real-time, the motor command, reaching a desired hip angular movement. Further, it leads to a safer system, reactive to patient's forces.

Obviously, the major achievement of this thesis is the *Solidworks* model of the prototype. The final assembly must be placed on the table and mobilizes the patient's leg thanks to a conic leg support. It adopts a flattened design, where the actuated linear guide rotates over an arc-shaped base. Attached to the driven carriage, the leg support travels laterally, back and forth, from the middle to the outside of the table, generating the abduction of the hip. Under the support, the force sensor collects the compression and traction forces in one direction, always normal to the moving leg. Additionally, to assure the stability of the device, a telescopic leg with an extended foot, fixed to the basis, stands firmly on the ground. Finally, to facilitate its transportation, the leg support can be detached from the base and the extensible leg folds, leading to a design that fits in a regular car trunk, weighing less than 20 kg.

9.2 Recommendations & next steps

This thesis suggests a comprehensive dynamic study of several mechanisms which results in a detailed and reflected mechanical design solution. Hence, it constitutes a strong basis for further development of this project, mainly the manufacture of the first physical prototype. However, there are definitely several areas of improvement that should be explored, down the road, to make this project real and successful in the medical world. We sincerely hope that this work will one day improve the quality of life of people who underwent a total hip replacement.

Validation of the *Robotran* model

The results obtained with *Robotran* were a key factor that conducted our choices and so, their impact on the final solution is major.

Hence, firstly, the simulation models should be upgraded by integrating the real physical characteristics (mass, inertia, dimensions, etc.) of the selected components. Thus, the correctness of the preliminary results could be assessed and the out-coming simulations would be closer to reality.

In a more distant future, the *Robotran* models should be validated with the physical prototype used in subjects, by collecting force thanks to the implemented transducer and comparing to the simulated forces.

Mechanical design

Several aspects could be pointed out concerning the design and assembly of the prototype.

As M-VITEC is a medical device, further focus should be given to the exterior design and ergonomics of the machine. In fact, its appearance should comfort the patient's mind, transmitting a feeling of safety and quality. For instance, a covering shell should be added above the engine and the base hiding as much as possible the moving parts.

Moreover, the placement of the connections and cables should be investigated to minimize their bulkiness and their interference with the moving components.

Finally, the selection of material concerning the components that need to be manufactured (i.e. base and leg support), deserves additional reflection in order to guarantee an adequate mechanical resistance, while reducing as much as possible the device's weight.

Instrumentation & control design

Unlike the initial perspectives, the final prototype solely acquires the patient's force in one direction. In fact, the integration of sensor capable of measuring also axial forces could lead to a safer device, but also allows a more complete validation of *Robotran* results.

Furthermore, in this thesis, no attention has been given to the control system and its design. This is probably the major aspect to investigate, for the progression of this project. A hardware solution should be selected, capable of collecting the sensor's output data. Control strategies must be explored and a design combining force and velocity (impedance control) must be investigated.

Finally, the emergency button is missing in the final prototype, so should be added such that the patient and the therapist can easily access it.

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Appendix A

Robotran models and results

Conceptual solutions: sol.1 to sol.9

This section provides detailed information about the *Robotran* analysis for all the solutions. For each of them, a *Robotran* model and a 3D representation are given, followed by four graphs gathering the results. They represent axial (with and without centrifugal effects) and normal forces in the hip. In addition, the motor requirements are shown, either as a force for linear actuated systems (solutions 1 to 7) or as a torque for direct-drive systems (solutions 8 and 9).

Solution 1

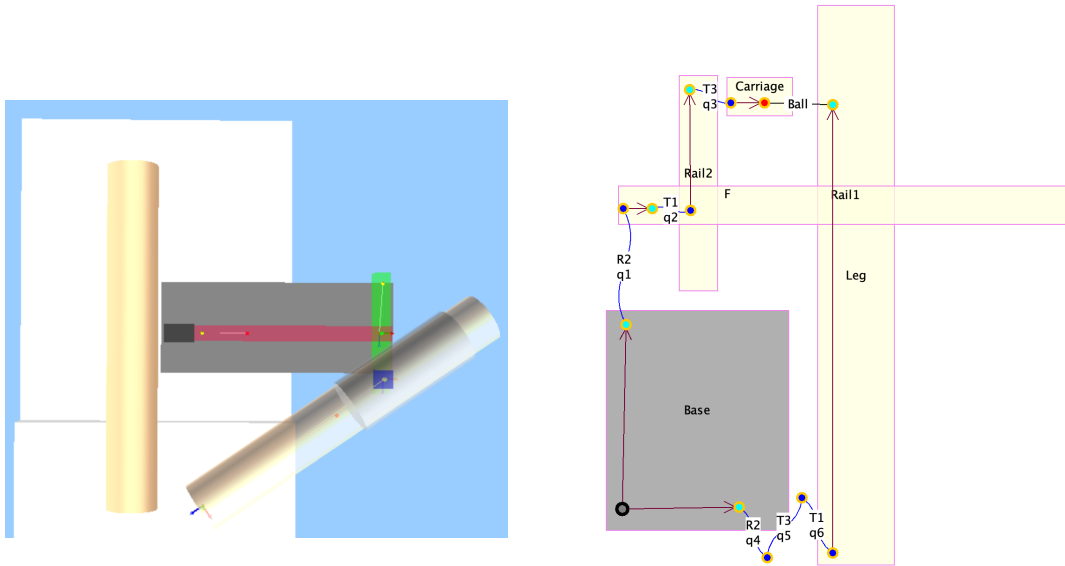
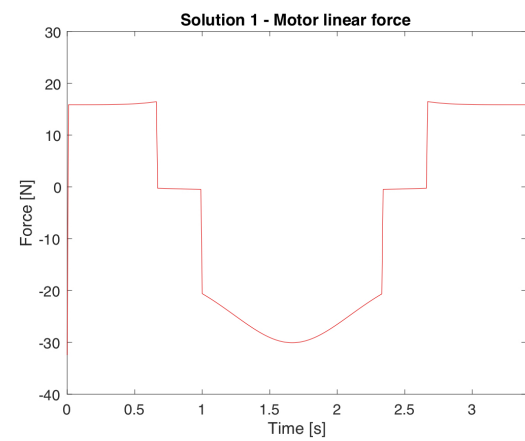
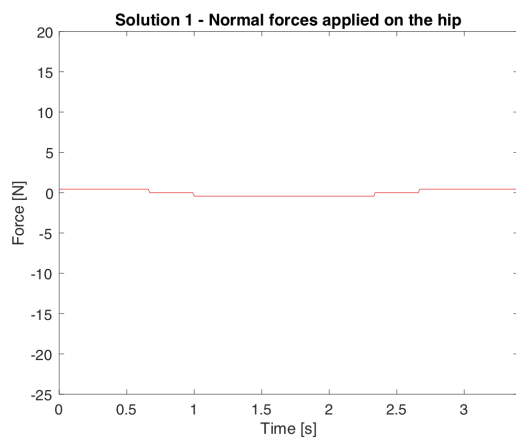
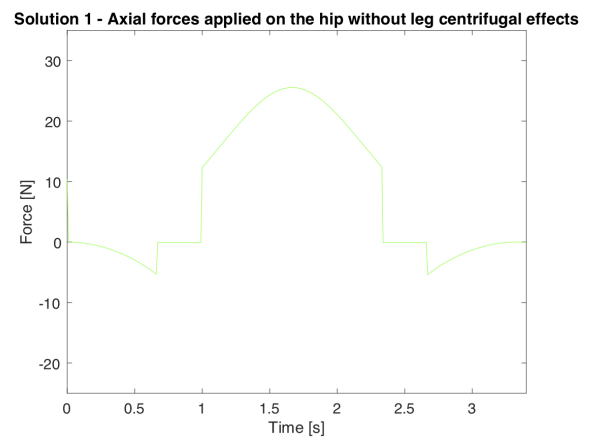
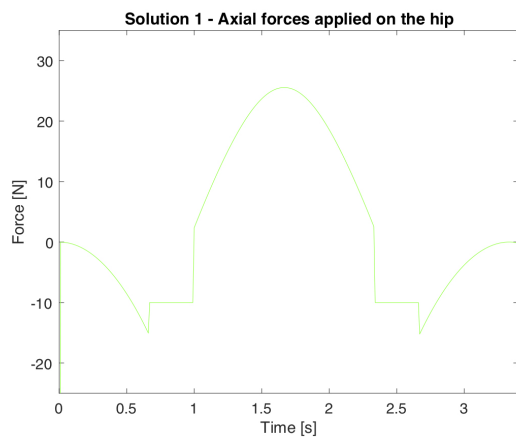


FIGURE A.1: Robotran model of solution 1



Solution 2

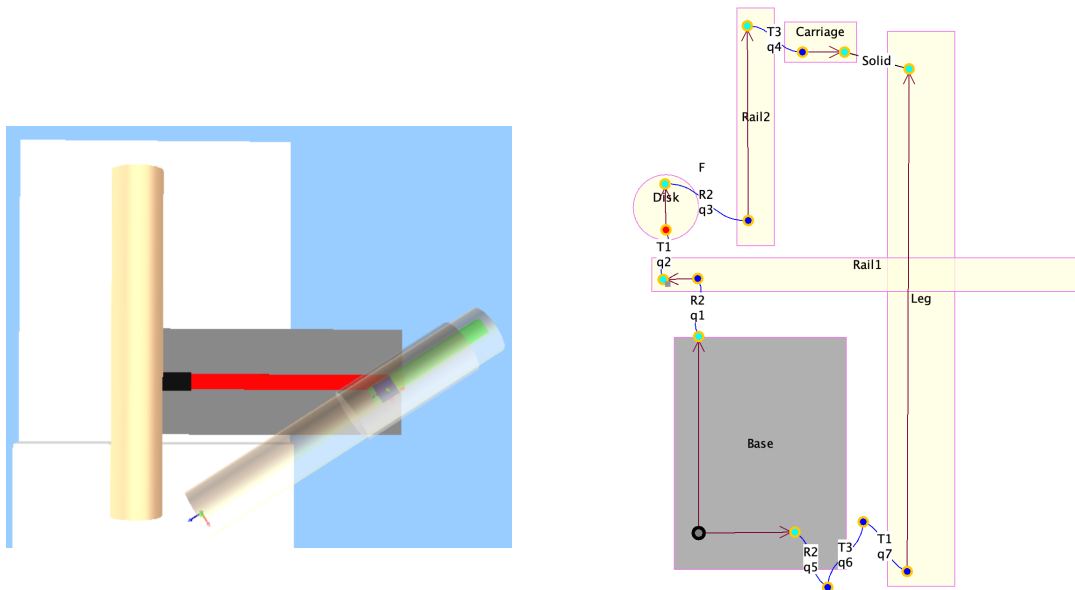
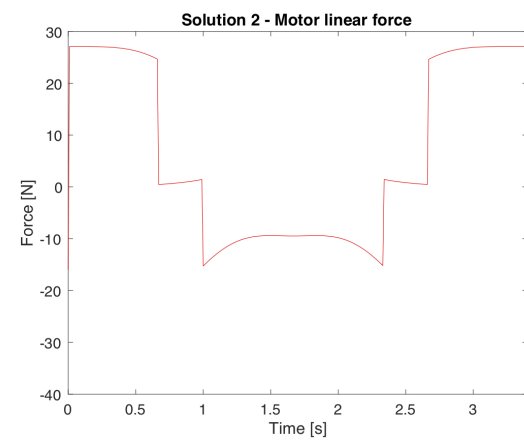
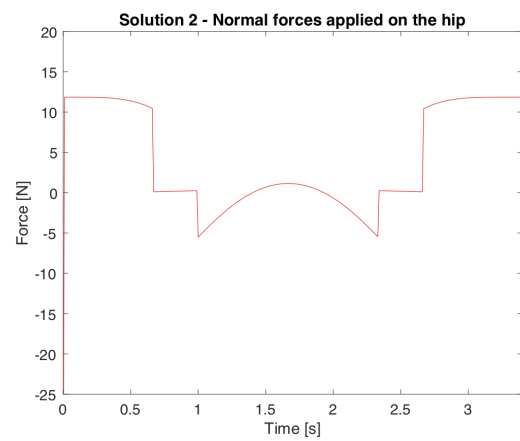
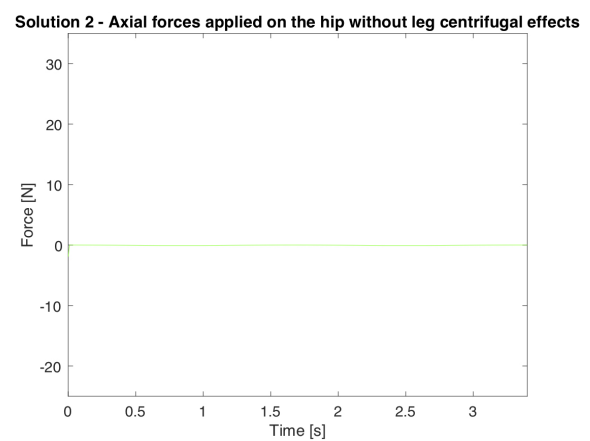
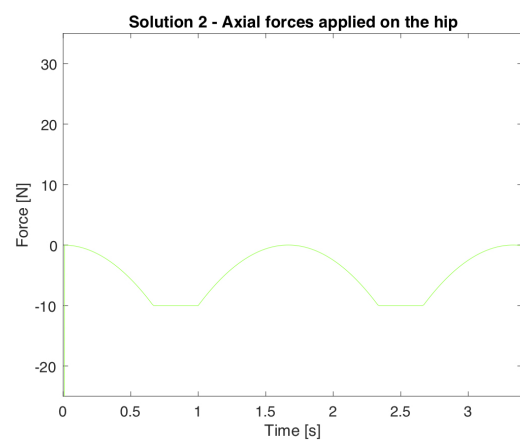


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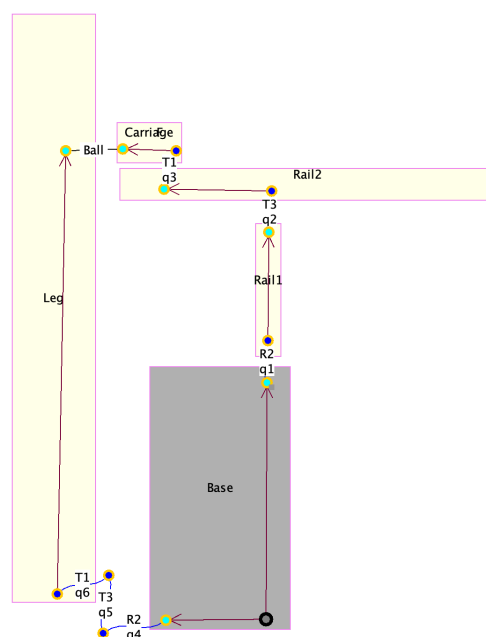
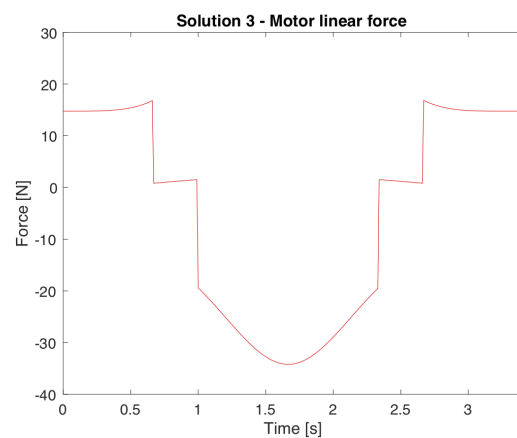
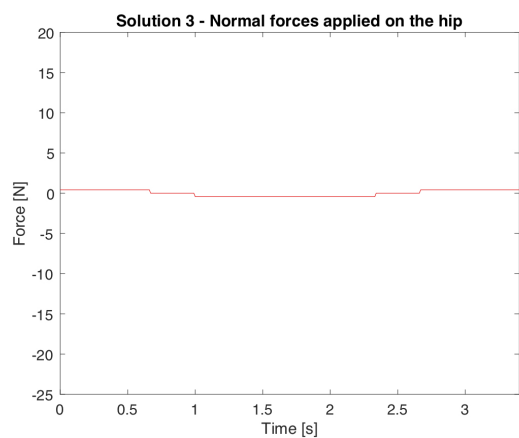
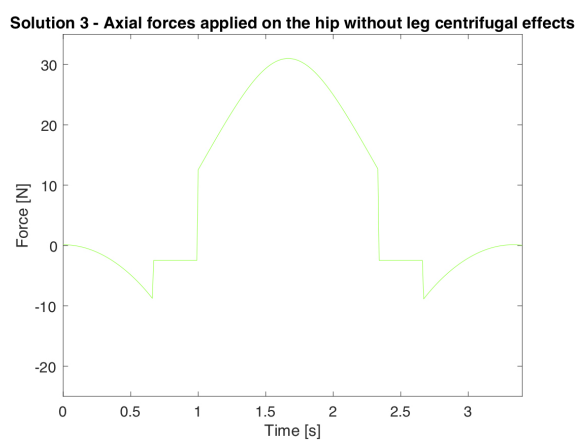
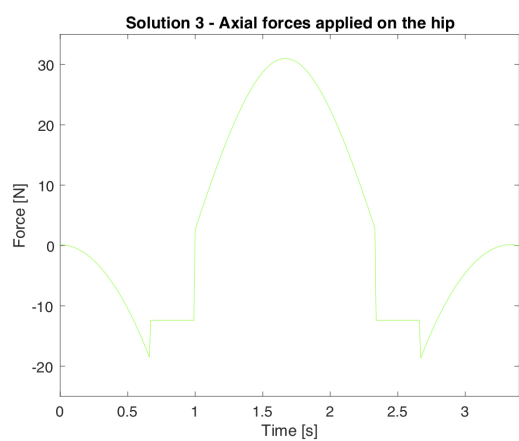


FIGURE A.3: Robotran model of solution 3



Solution 4

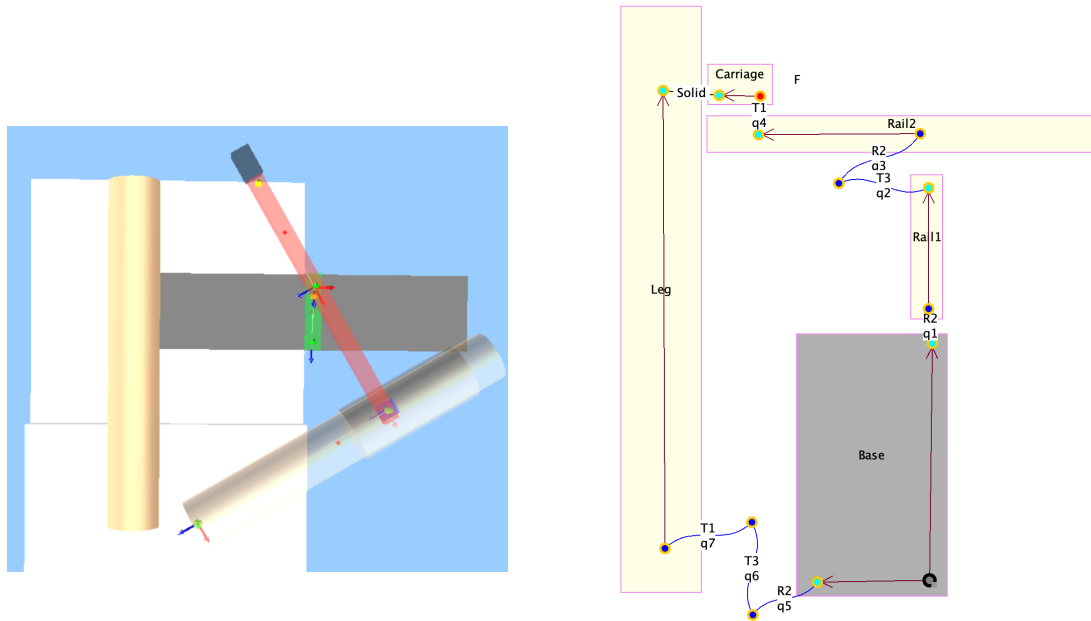
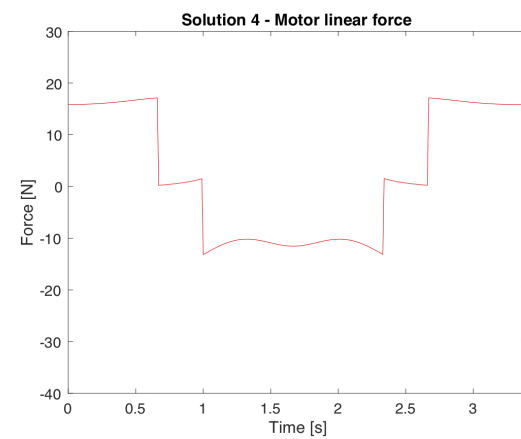
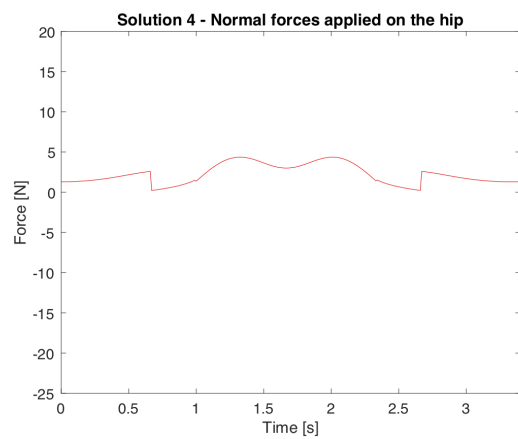
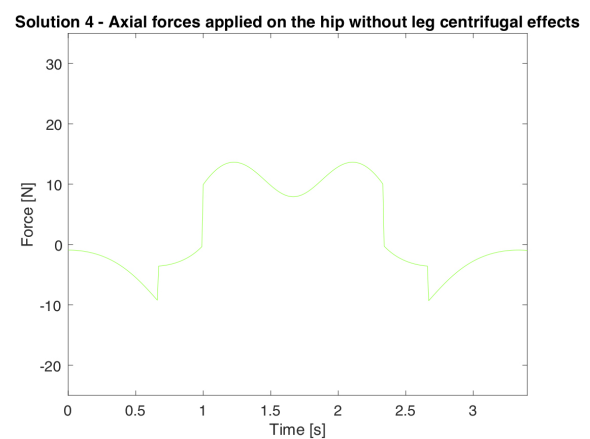
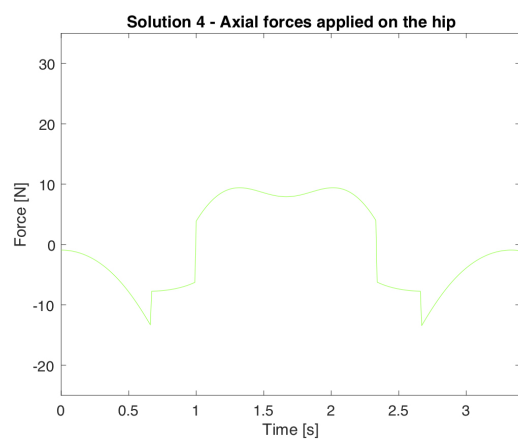


FIGURE A.4: Robotran model of solution 4



Solution 5

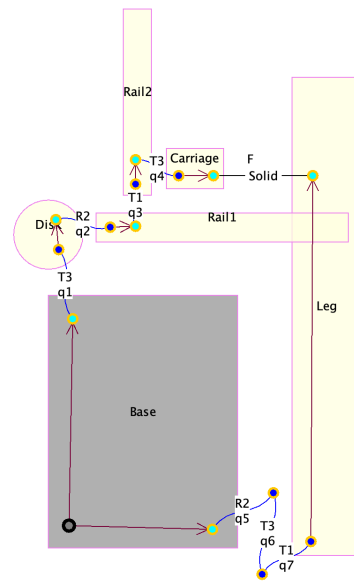
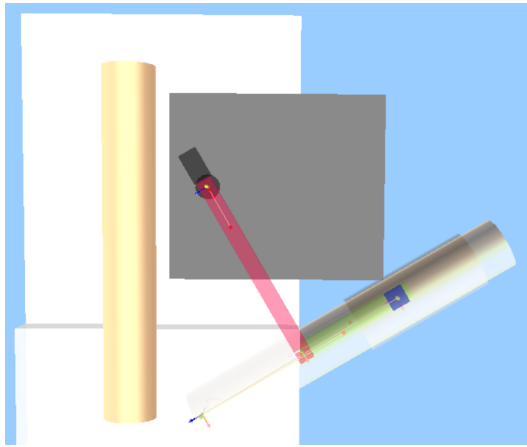
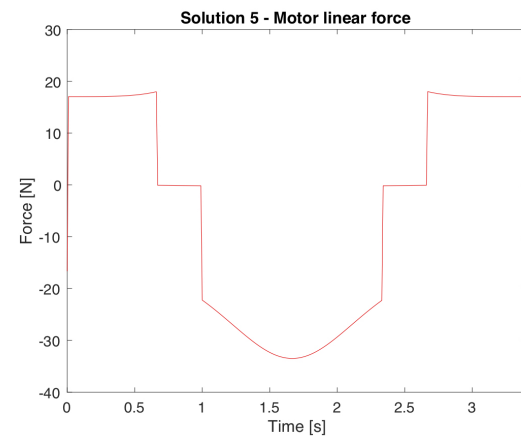
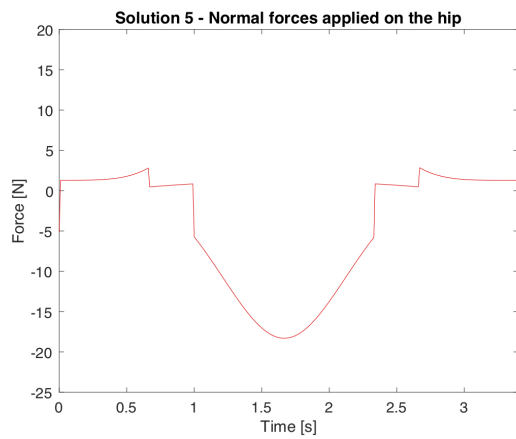
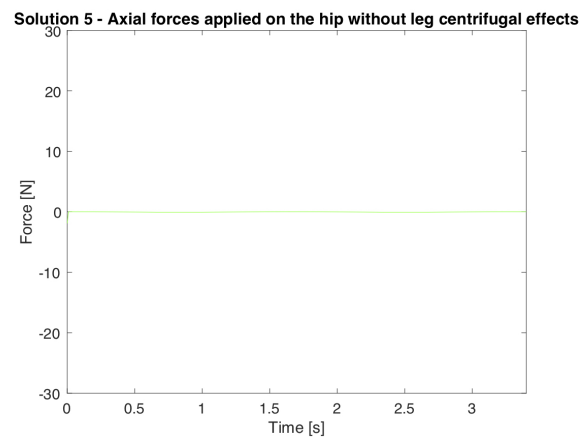
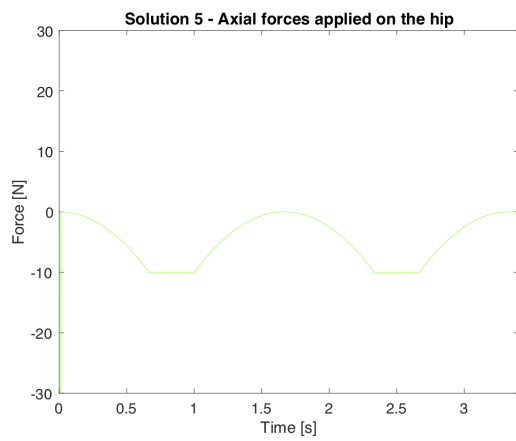


FIGURE A.5: Robotran model of solution 5



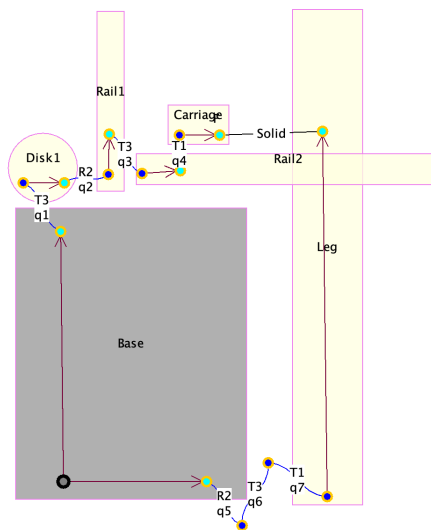
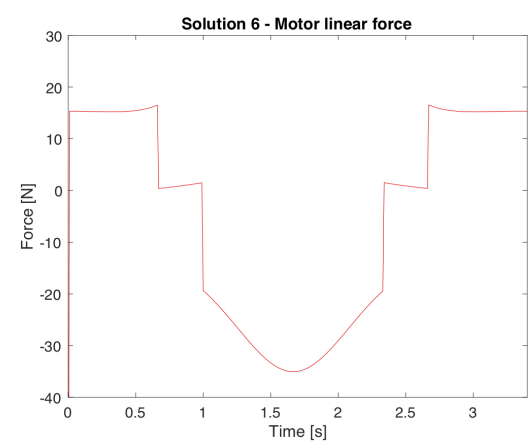
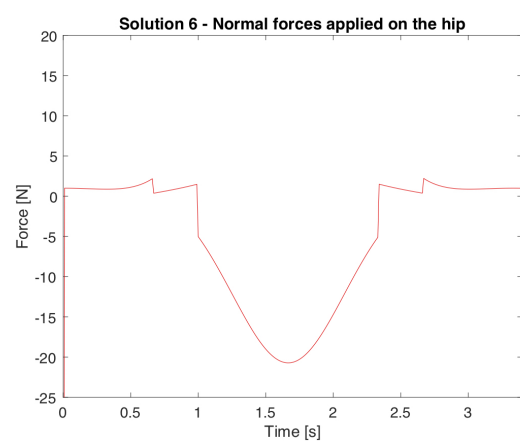
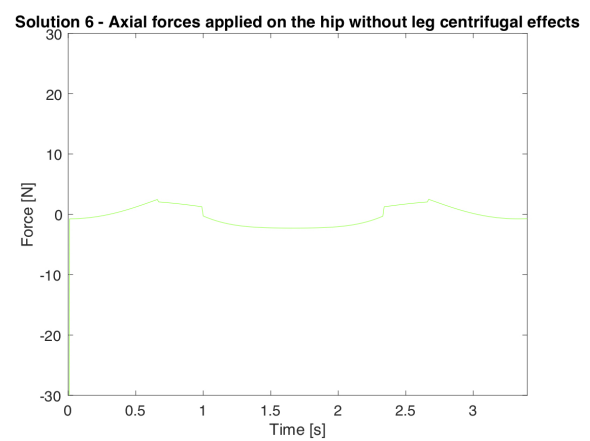
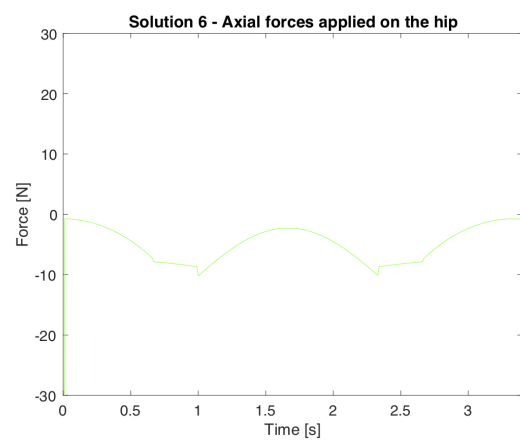


FIGURE A.6: Robotran model of solution 6



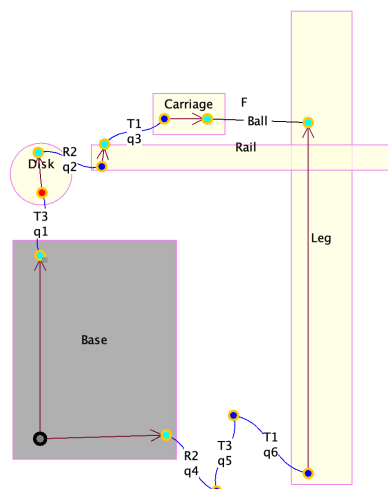
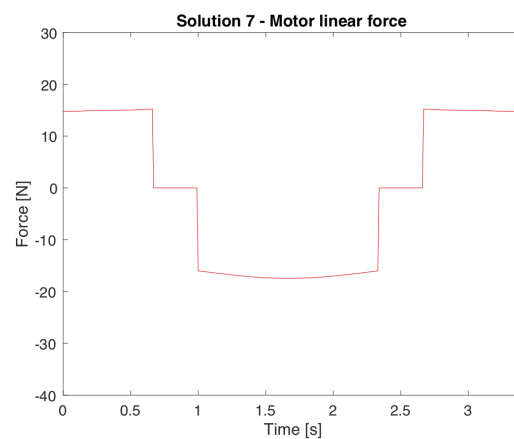
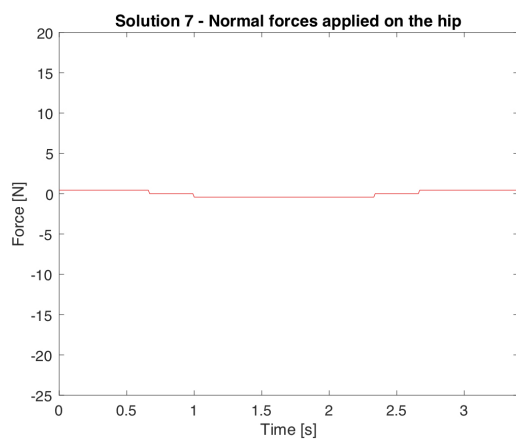
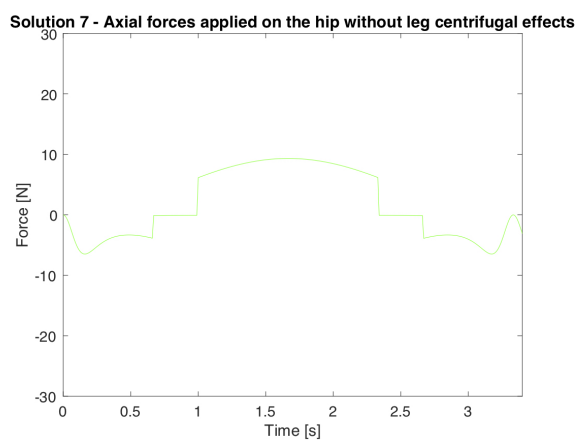
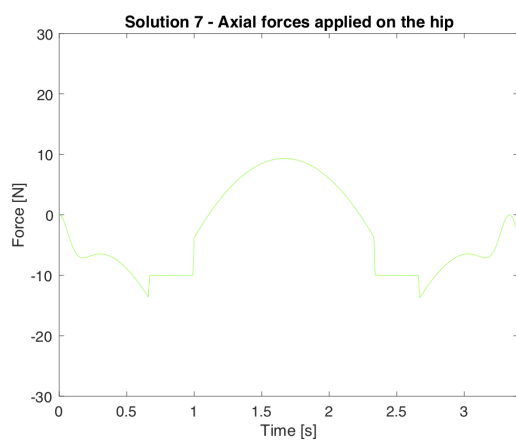


FIGURE A.7: Robotran model of solution 7



Solution 8

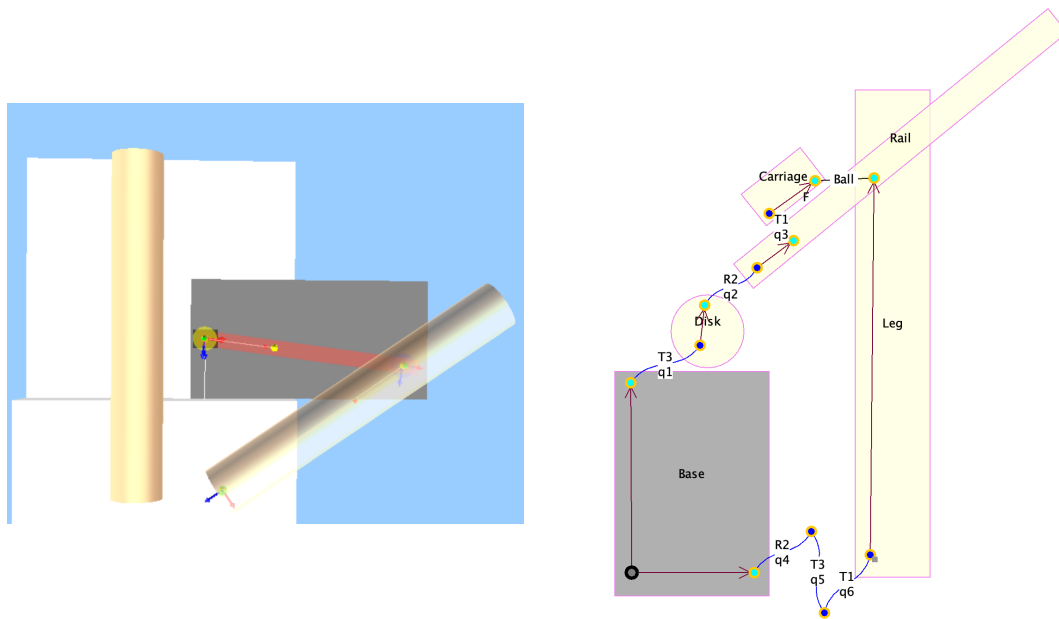
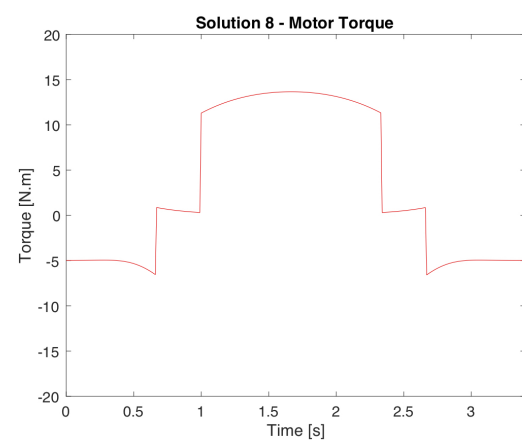
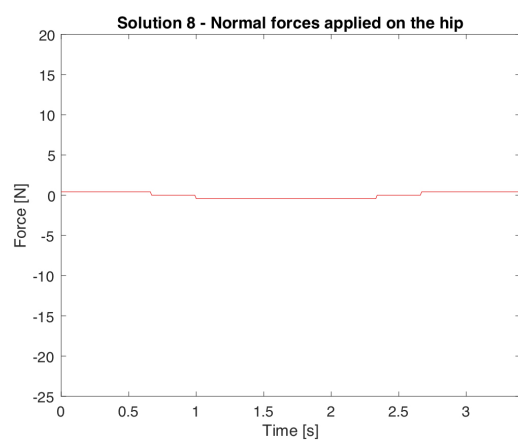
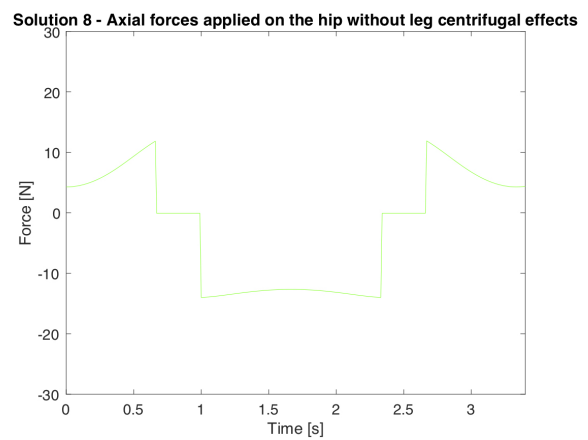
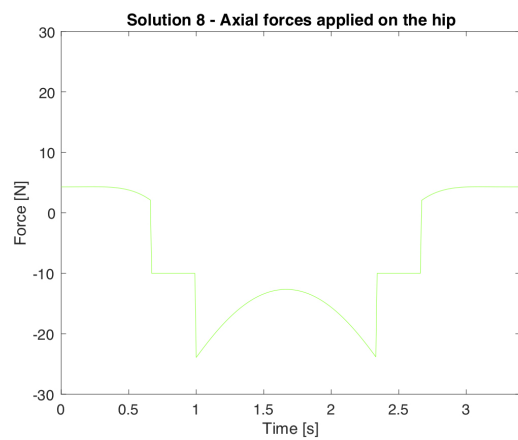


FIGURE A.8: Robotran model of solution 8



Solution 9

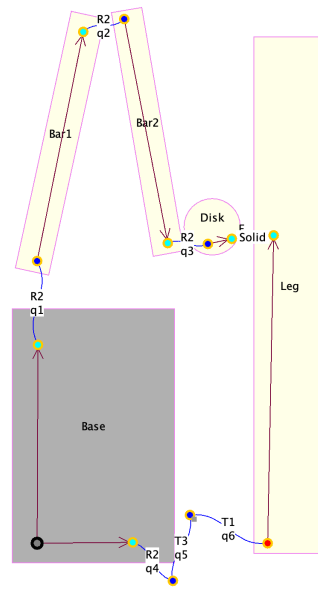
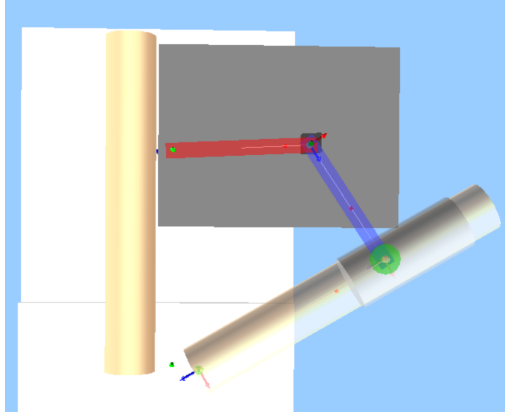
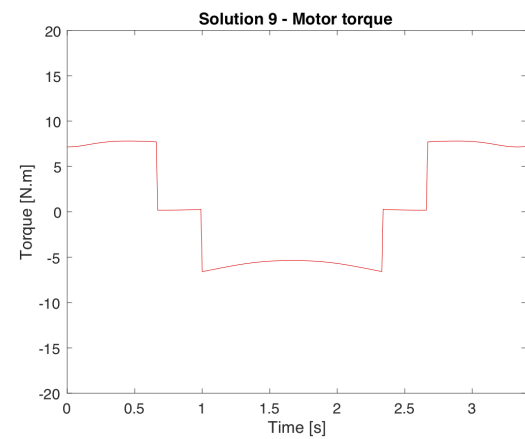
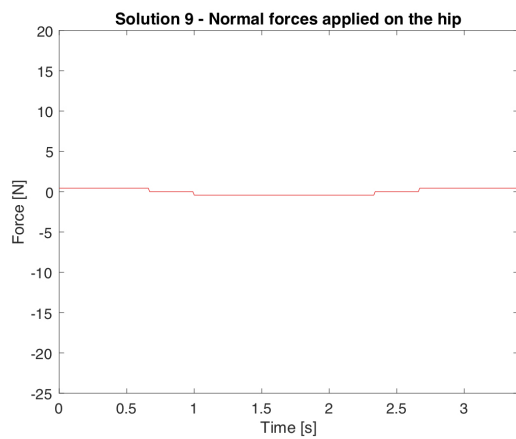
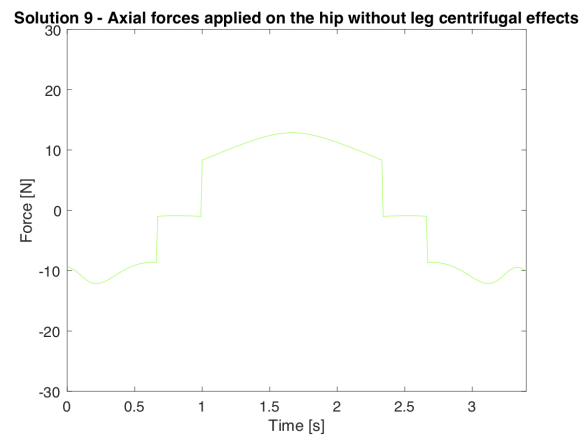
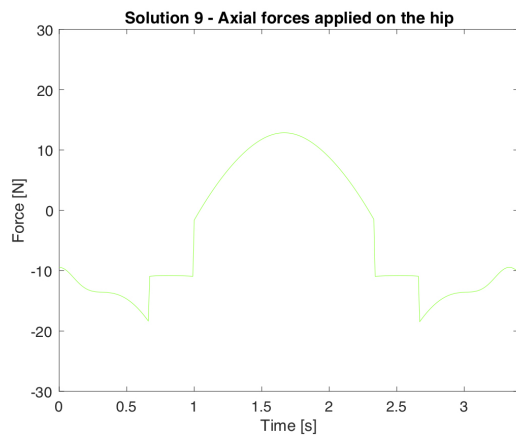


FIGURE A.9: Robotran model of solution 9



Alternative solutions

This section presents the additional solutions thought during the conceptual design. In fact, all nine solutions considered in this conceptual design have an "externalized" equivalent, composed of the same kinematics but the device's engine and basis are positioned next to the bed/table rather than on the bed/table. Thus, the device would be hold by a trolley anchored in the floor and equipped with wheels enabling transportation.

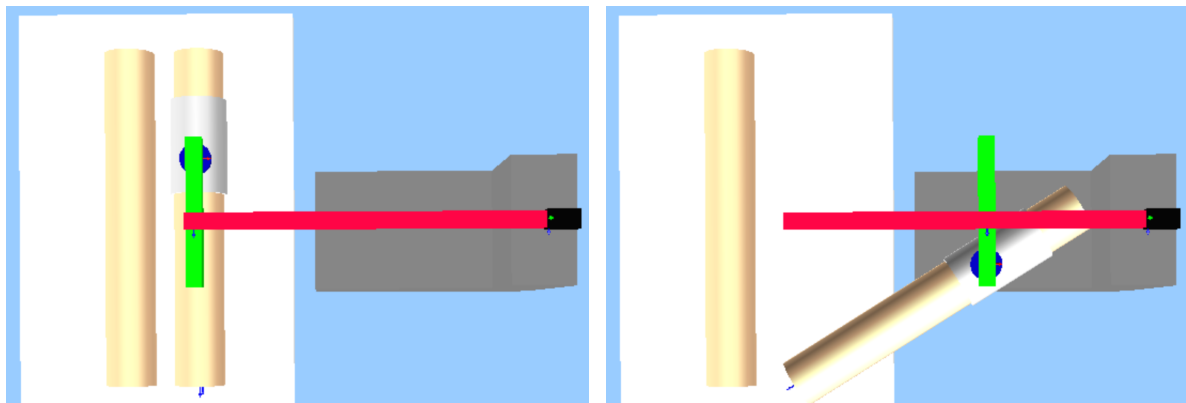


FIGURE A.10: Alternative for solution 1

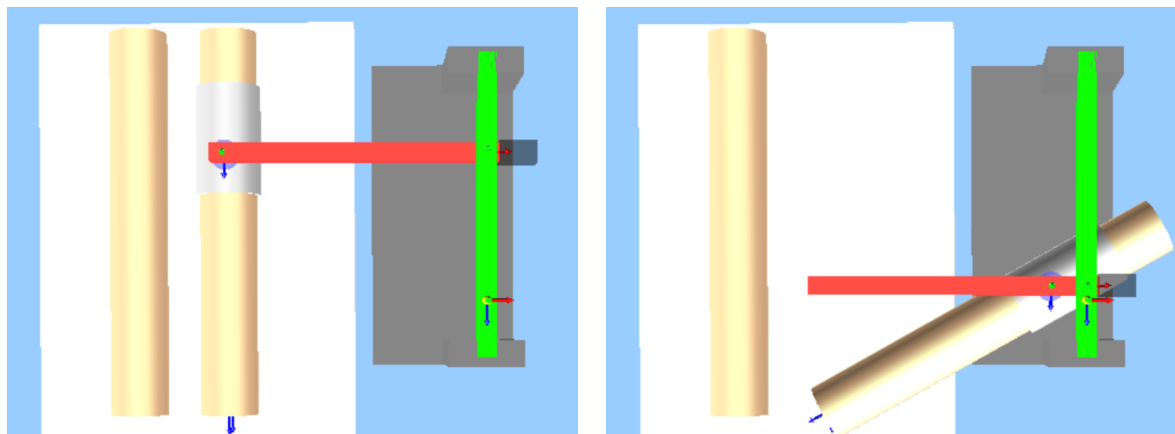


FIGURE A.11: Alternative for solution 3

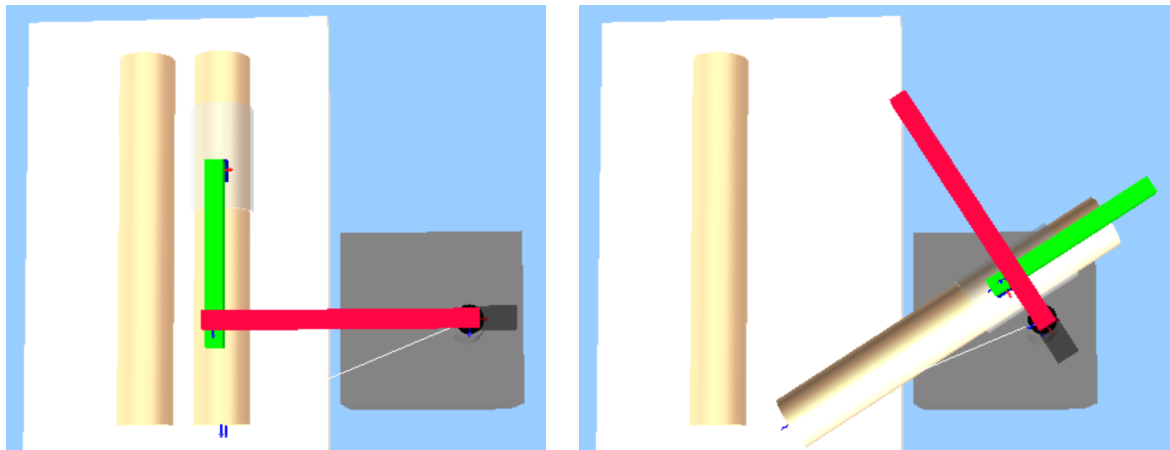


FIGURE A.12: Alternative for solution 5

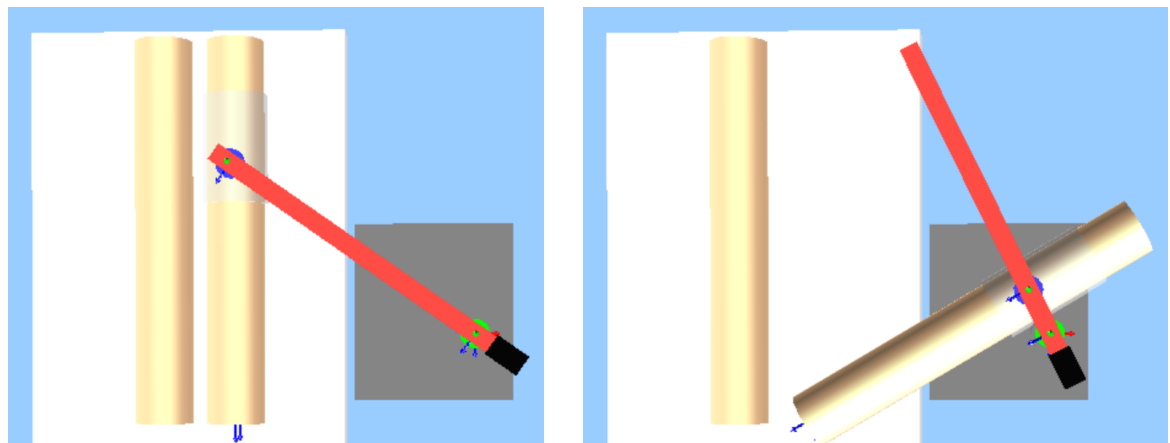


FIGURE A.13: Alternative for solution 7

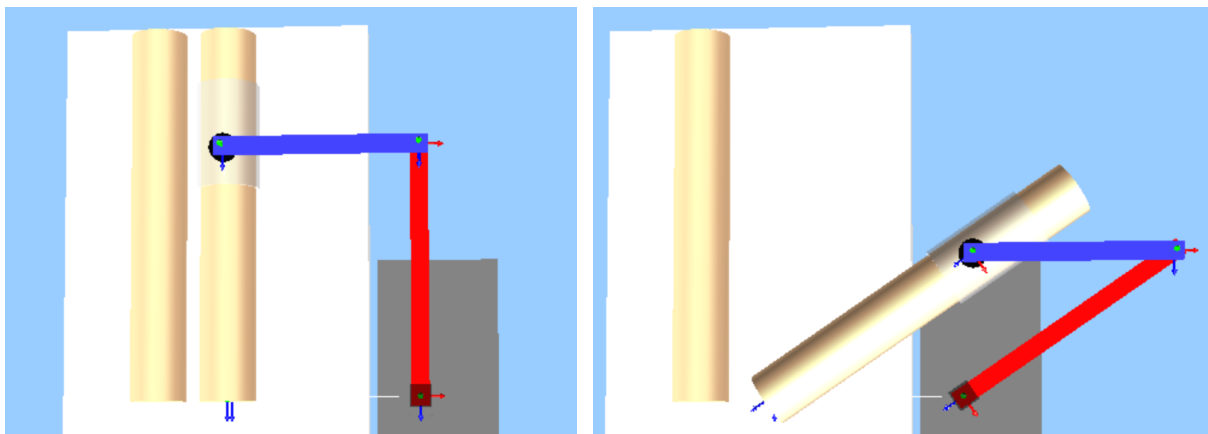


FIGURE A.14: Alternative for solution 9

Appendix B

Risk analysis

This preliminary analysis is a simplified "Failure mode and effects analysis" (FMEA) done in the frame of LMECA2855 course thanks to *Matrix Requirements* software [95]. For each identified risk when operating with the device, the following points were defined:

- **Hazard:** Briefly outline the device function or action that makes it potentially dangerous.
- **Sequence of events:** Summary description of the actions leading to the hazardous situation.
- **Hazardous situation:** Specific action that might harm the patient.
- **Harm:** The consequence on the patient health.
- **Probability of occurrence:** How often the hazardous situation happens (from 1 to 5).
- **Probability of harm:** If the hazardous situation happens, how often it harms the patient (from 1 to 5).
- **Severity:** If the patient is harmed, how severe is the impact on his health (from 1 to 5).
- **RPN:** Risk priority number. It is obtained multiplying the three previous numbers (from 1 to 125). If it is above a preset cutoff value (i.e. 15), it is highly recommended to take risk measures lowering the RPN.

Item	Hazard	Sequences of events	Hazardous situation	Harm	Prob. of occurrence	Prob. of harm	Severity	RPN
RISK 1: Material	The patient might have allergies to some materials.	Device is in contact with the patient.	Patient has an allergic reaction with the materials that are in contact with him.	Patient has an irritation/allergic reaction.	1	1	3	3
RISK 2: Maladjusted fixations	The device features fixations for the patient's lower limb that can be adapted to different morphologies.	The patient inserts his injured leg into the fixations. Therapist secures the leg badly.	There is some backlash between the patient's leg and the fixations.	Irritation, wrong measurements, instability.	2	2	2	8
RISK 3: Movement amplitudes	The device is capable of reaching amplitudes of abduction / adduction that the patient may not be able to achieve.	The patient places his injured leg into the fixations. The therapist calibrates the device so that the amplitude of the movements performed are attainable by the patient, and appropriate for revalidation.	Excessive abduction /adduction because of bad calibration.	Dislocation of the hip, tendon or muscle tearing.	1	3	4	12
RISK 4: Cleanliness	The fixations will be in contact with a variety of patients.	Multiple patients use the device over time. Their hygiene might be insufficient.	The parts of the device in contact with the patients are contaminated.	Propagation of diseases or fungi. Cutaneous conditions.	3	1	3	9
RISK 5: Measurement sensors	The sensors used in the device are not fully precise.	The patient is fastened to the device. The therapist starts the revalidation session. The sensors return wrong measurements.	The controller receives corrupted inputs.	The device performs movements that could harm the patient, provoking possible dislocations/injuries.	2	2	4	16
RISK 6: Software	Software controls the device.	Start of the revalidation session.	Malfunction of the device	From wrong treatment to patient injuries.	1	2	3	6
RISK 7: Broken Fixations	Fixations hold the leg steady.	Patient places his injured leg into the fixations. Start of the revalidation session. The force applied by the patient damages the fixations.	The fixations break under the forces applied by the patient.	Wrong movement, wrong leg position, possible injuries.	1	1	3	3

RISK 8: Emergency button	There is an emergency button to stop the device immediately.	The patient wants to stop the device. The patient pushes on the emergency button.	The emergency button does not stop the operation of the device.	Possible injuries or hip dislocation.	1	3	4	12
RISK 9: Maintenance	Maintenance has to be performed regularly to ensure proper operation of the device.	Maintenance is long overdue.	Loss or deterioration of the device's function (non-isokinetic treatment)	Bad treatment, shaken lower limb.	2	2	1	4
RISK 10: External obstruction	The device has sliding parts.	Therapist session begins. The device performs abduction/adduction movements	Bed sheets or a clothing come into a sliding part and prevent the device from doing the movement of abduction and adduction.	Can damage the device (and eventually harm the patient with sudden movements)	2	1	3	6
RISK 11: Magnitude of the force	The device is capable of applying forces that the patient might not be able to handle.	The patient places his injured leg into the fixations. The therapist calibrates the device so that the forces applied are bearable by the patient, and appropriate for revalidation.	Excessive forces because of bad calibration.	Dislocation of the hip, tendon or muscle tearing.	1	3	4	12
RISK 12: Stability	The device needs to be secured to be stable.	The device is placed on the hospital or physiotherapist bed. The clamps are wrongly fixed on the bed by the therapist. The leg is fastened on the leg support. The device starts performing movements.	The device falls on the floor.	Possible injuries or hip dislocation.	2	2	3	12

FIGURE B.1: LMECA2855 risk analysis

Appendix C

Igus quote

The Appendix contains the price quote of the actuation system suitable for our application proposed by *Igus*. The second page contains three propositions, the first include the following items:

- Toothed belt axis
- Belt
- Carriage
- Stepper motor
- Encoder
- Two inductive sensors
- Connecting cables

The second is similar to the first but one extra connecting cable was added. The third only concerns the motor controller. Prices are found next to each proposition.

Note that theses quotes consider a slightly longer belt-driven axis (900 mm). In reality, the final choice of stroke was 800 mm. Furthermore, the two inductive sensors are not used final design, since we decided to replace it by one micro-switch integrated in the carriage. Finally, the controller properties be investigated to make sure it can be configured according to our desires.



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Date : 12.06.2019
Votre numéro client : 6949345
Votre demande :
Votre contact : Simon Bodart
Téléphone : +32 476985201
Fax : +32 3 313 79 91
E-mail : sbodart@igus.net

Numéro d'offre 19-092311-BRG-BE-SBO [Numéro d'offre à spécifier sur votre commande]

Monsieur Vasco José de Melo,

Nous vous remercions pour l'intérêt que vous portez à notre gamme de produits. Veuillez trouver ci-dessous notre offre de prix:

Pour toutes informations complémentaires notamment concernant le montage de nos produits, vous trouverez des précisions sur nos catalogues et sur notre site internet www.igus.be. Nous sommes également à votre entière disposition au +32 3 330 13 60.

Pour les commandes, vous pouvez simplement nous envoyer un courriel à commandes@igus.be

Sincères salutations
igus® B.V.B.A.

Simon Bodart
Outsidesales BRG / Product Manager iglidur

Pos.	Référence	Quantité	Unité	Prix Unitaire	Prix total HT
1.000	ZLW-1040S-G0BW0-D0A1C-AN0F1AA-900 drylin® ZLW-1040-standard toothed belt axis, Profile: hard-anodised, PU toothed belt with steel cord, white, carriage length 100 mm, end block made of plastic, left version, NEMA23XL stepper motor, encoder, 3.5 Nm, 2 x initiator kit, inductive, 24 V, PNP NC, with bracket, 3 meter cable with straight connector, mounting 0°, Stroke 900 mm, system length (l+stroke) 1104 mm Cette position comporte une remise de 15,0%.	1	1 Pièce(s)	EUR 823,54 / 1	EUR 823,54
2.000	ZLW-1040S-G0BW0-D0A1C-AN1F1AA-900 Alternative drylin® ZLW-1040-standard toothed belt axis, Profile: hard-anodised, PU toothed belt with steel cord, white, carriage length 100 mm, end block made of plastic, left version, NEMA23XL stepper motor, encoder, 3.5 Nm, 3 metre cable with straight connector, 2 x initiator kit, inductive, 24 V, PNP NC, with bracket, 3 meter cable with straight connector, mounting 0°, Stroke 900 mm, system length (l+stroke) 1104 mm Cette position comporte une remise de 15,0%.	1	1 Pièce(s)	EUR 892,47 / 1	(EUR 892,47)
3.000	D1 dryve drylin E Cette position comporte une remise de 15,0%.	1	1 Pièce(s)	EUR 287,51 / 1	EUR 287,51
Total des positions de l'offre en EUR HT					EUR 1.111,05

Appendix D

Prototype components

This appendix provides additional information about the 145 components composing the prototype. For each part, several details are provided, such as the brand, the reference and the price. They are divided in seven categories:

- Electronic components
- Linear system components
- Plastic components
- Bearings, spacers, shaft and pivots
- Nuts, screws and washer
- Custom components (to be manufactured at *UCLouvain* or replaced by compatible prefab part)
- Other components

In addition, the total price in euros was computed (considering the components for which the price was available).

Category	Component	Num.	Brand	Supplier	Ref.	Qty	Price
ELECTRONIC COMPONENTS	Radial Encoder Kit 9 sleeve	8	CUI	Mouser	AMT102-2048-i6000-w	1	20.63
	Tension/compression force transducer	33	TECSIS	TECSIS	F2220	1	-
	Snap action switch	59	Marquardt	Mouser	1058.0351	2	3.44
	Snap action switch with roller arm	46	Würth Elektronik	RS Components	463093370402	1	3.85
LINEAR SYSTEM COMPONENTS	ZLW-Toothed belt axis (carriage included)	14,15	Igus	Igus	ZLW-1040-02-S-100-R	1	436.50
	Stepper motor NEMA23XL	-	Igus	Igus	MOT-AN-S-060-035-060-M-C-AAAC	1	287.17
	ZLW – NEMA motor flange	-	Igus	Igus	MF-1040-NEMA23-S	1	55.64
	Coupling	-	Igus	Igus	COU-AR-K-080-080-32-32-B-AAAA	1	29.90
	Slot nuts	13	Igus	Igus	NOR-20602	6	1.14
	Pillow block with manual clamp	51	Igus	Igus	WHKA-10	1	16.01
	Miniature Linear Guide	28,31	MISUMI	MISUMI	SSEB6-25	1	26.16
PLASTIC COMPONENTS	Device's Base	5	-	-	-	1	-
	Leg Support	40	-	-	-	1	-
	Rubber impact buffer	62	norelem	norelem	26108-02601560	2	0.91
	Low rebound urethane bumper	39	MISUMI	MISUMI	AXFAN-D14-L12-V102	1	4.95
	Plastic knob	19	MISUMI	MISUMI	NKSM5-15	1	1.29
	Plastic knob	-	MISUMI	MISUMI	NKSM6-10	1	1.20
BEARINGS, SPACERS, SHAFT AND PIVOTS	Bearing with housing-Standard	4	MISUMI	MISUMI	SBARA626ZZ	1	20.11
	Bearing with housing-Compact	22	MISUMI	MISUMI	BGTN626ZZ	1	19.44
	Flanged deep groove bearing	11	MISUMI	MISUMI	SFL696ZZ	2	5.00
	Precision pivot pin	10	MISUMI	MISUMI	SCLBK6-10-FC8	2	10.28
	Precision pivot pin	24	MISUMI	MISUMI	CLBK6-10.5-FC5	1	7.4
	Cantilever shaft	6	MISUMI	MISUMI	FXCA6-2-F15-G9	1	18.40

Category	Component	Num.	Brand	Supplier	Ref.	Qty	Price
	Bearing spacer for inner ring	12	MISUMI	MISUMI	CLBPS6-9-6.2	2	5.40
	Bearing spacer for inner ring	3	MISUMI	MISUMI	CLBPS6-9-5	1	5.40
	Bearing spacer for inner ring	26	MISUMI	MISUMI	CLBPS6-9-4	1	5.40
NUTS, SCREWS AND WASHER	Flat head cap screw	7	MISUMI	MISUMI	SFB4-12	4	0.55
	Flat head cap screw	23,44	MISUMI	MISUMI	SFB4-8	11	0.54
	Flat head cap screw	21	MISUMI	MISUMI	SFB5-10	2	0.56
	Flat head cap screw	16	MISUMI	MISUMI	SFB5- 20	4	0.57
	Socket head cap screw	-	MISUMI	MISUMI	SCB5-25	2	0.80
	Flat head cap screw	50	MISUMI	MISUMI	SFB5- 30	4	0.58
	Socket head cap screw	41	MISUMI	MISUMI	SCB6-18	4	0.80
	Socket head cap screw	-	MISUMI	MISUMI	SCB6-30	4	0.88
	Ultra low head sap screw	47	MISUMI	MISUMI	CBSTSR2-10	2	2.60
	Ultra low head cap screw	25	MISUMI	MISUMI	CBSTSR2.5-6	2	2.60
	Ultra low head cap screw	35	MISUMI	MISUMI	CBSTSR3-8	6	3.15
	Ultra low head cap screw	-	MISUMI	MISUMI	CBSTSR4-16	4	2.94
	Ultra low head cap screw	60,61	MISUMI	MISUMI	CBSTSR5-25	10	2.80
	Extra low head cap screw	58	MISUMI	MISUMI	CBSTS6-20	1	3.15
	Ultra low head screw	-	MISUMI	MISUMI	CBSTSE5-30	2	2.20
	Low head cap screw	-	MISUMI	MISUMI	CBSST5-20	1	0.38
	Low head cap screw	-	MISUMI	MISUMI	CBSST5-16	1	0.38
	Low small head cap screw	9	NISSAN SCREW	MISUMI	CSHSMT-SUS-M3-14	2	0.93
	Flat head bolt with hex socket	30	SUNCO	MISUMI	CSHC5-ST-M2-8	2	0.73
	Hex socket set screws with soft tip	42	MISUMI	MISUMI	SBPP6-30	2	1.95
	Hex socket set screws with soft tip	53	MISUMI	MISUMI	SBPP6-20	1	1.95

Category	Component	Num.	Brand	Supplier	Ref.	Qty	Price
	Cross recessed pan head screw	48	MISUMI	MISUMI	NKJ5-12	1	0.86
	Cross recessed pan head screw	-	MISUMI	MISUMI	NKJ6-12	5	0.87
	Compact nut	1,27,45	MISUMI	MISUMI	SNTRC5	5	2.00
	Hex nut	55	MISUMI	MISUMI	SLBNR6	1	0.51
	Hex nut	-	MISUMI	MISUMI	LBNR5	2	0.28
	Hex nut	36	MISUMI	MISUMI	LBNR10	1	0.47
	Flat washer	37	MISUMI	MISUMI	PWF10	1	0.19
CUSTOM COMPONENTS	Telescopic folding leg	-	-	-	-	1	-
	Flat bar switch mounting plate	63	-	-	-	1	-
	Flat bar force sensor mounting plate 1	29	-	-	-	1	-
	Flat bar force sensor mounting plate 2	34	-	-	-	1	-
	Flat bar switch mounting plate	43	-	-	-	1	-
	L-shaped switch mounting plate	-	-	-	-	1	-
	L-shaped bumper mounting plate	-	-	-	-	1	-
	L-shaped carriage mounting plate	20	-	-	-	1	-
	Mounting flanged block	57	-	-	-	1	-
	Wedged-shaped plate	54	-	-	-	1	-
	Mounting rod	52	-	-	-	1	-
	U-shaped bearings holder	2	-	-	-	1	-
	U-shaped block (leg support holder)	32	-	-	-	1	-
	U-shaped plate fixed to the carriage	17	-	-	-	1	-
	U-shaped plate covering the carriage	18	-	-	-	1	-
	End of stroke plate	49	-	-	-	1	-

Category	Component	Num.	Brand	Supplier	Ref.	Qty	Price
OTHER COMPONENTS	8 inch folding shelf bracket	-	DMSL	Amazon	B07V7TFK16 (ASIN)	1	22.61
	Stainless steel handles	-	MISUMI	MISUMI	UWANDC115	2	9.50
	Vertical handle flange base	56	MISUMI	MISUMI	MC04-S2	1	19.95
	Ball transfer unit	38	IGUCHI KIKO	MISUMI	IK-19SNM	1	87.73
					TOTAL	145	1368.07

Appendix E

Robotran parameters

Patient characteristics

- Height: 2 m
- Weight: 100 kg
- Body density : $1062 \text{ kg}/\text{m}^3$

Mechanical components

Mass estimation:

- Leg support: 2 kg
- Motor : 1.5 kg
- Linear guide and rotary joints: Mass density = $1000 \text{ kg}/\text{m}^3$

Height estimation:

Components	Height [cm]
Motorized linear guide	6
Non-motorized linear guide	5
Revolute joint	2
Leg support	3
Rod	2

TABLE E.1: Reference components heights used for a preliminary device height estimation [60][61][96][97]

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