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Biomedical Engineering

BMS3686 Principles of Rehabilitation Engineering

Coursework A

Analyze of human movement and solving biomechanics of the musculoskeletal system

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Introduction

The analyse of human movement and its biomechanics represents an important task for the rehabilitation of many medical conditions. In addition, many medical fields have begun to examine the utility of optimization techniques to assist the understanding of the biomechanics (Rapoff et al., n.d.). However, understanding the biomechanics of the human body motion involves a clear knowledge of the musculoskeletal system and its inversive dynamics.

Musculoskeletal system

The human musculoskeletal system is an organ system that provides the support, the ability to movement and stability. It is composed by two fundamental systems, the skeletal and the muscular system, that work together to support the body weight.

The skeletal system is a key component in the biomechanics and the body, as it provides support and protection, possibilities the body movements, maintains the minerals and fat and is the local where blood cells are produced. Besides bones, the skeletal system also possesses cartilage. There is tree types of cartilage: hyaline cartilage, composed by chondrocytes, located at the end of the long bones, and provides a slippery surface; elastic cartilage, that is highly bendable and elastic and fibrocartilage, resistant to compression and tensions seen in the knee menisci and or in intervertebral discs.

Bones are the main component in the skeletal system and in total it exists 206 bones in the human body. There are two basic types of bone tissue, cortical bone that are dense and homogeneous and trabecular bone that are more porous and have open spaces. Cortical bones are highly organized and strong, mainly consisted by haversian system. Trabecular bone is a lighter and less dense bone tissue, they consist of plates that connect to each other to form a mesh for the red bone marrow and form lines of stress to provide maximum strength and can remodel in case the direction of stress changes.

The bottom part of the human body is where we find the bones that compose the inferior limbs, the main supporters of the body weight and responsible for our stability and movement. The lower limbs consist of pelvis, legs and feet. The bones that composed these limbs are the hip bone, femur, kneecap, tibia, and the feet bones (tarsal, metatarsal, and phalanges).

The muscle is what allows the human body to movement. We move because the muscles pull the bones, but this movement could not happen if we did not had joints. Joints is the place where two bones, usually moving, come in contact and, accordingly with the function and the bones that are related, they can contain more or less fibrous joints and a larger or smaller range of movement. There are three types of joints. For movements such as walking, the joints that can be seen are mainly synovial joints ,that contain a synovial liquid and allow a large range of movement between the bone that articulates the allowed movement. This can be seen in the hip by a ball-and-socket joint that allows multi-axial movements, the pivot joints in knee meniscus that only allows bi-axial movement.

The muscles are a highly specialized tissue that do contraction or shortening work. This process is done by a metabolic rate that consumes energy through the nutrients molecules. The main properties of the muscles are the contractability, originated by muscles fibre moved by the *epimysium,* that generates enough force to make the structures that are attached to move. The muscles are also highly elastic and excitable to stimuli done by either the nervous system or hormones (Seeley, Stephens and Tate, 2003).

Muscle and leg

The muscles that compose the leg contain a large variety of different muscles. While some are long and strong muscles and produce the majority of the work involved during movement, there are also small muscles responsible for helping the stability of the joints, the rotation and facilitate other movements. The quadriceps are the strongest and leanest of all muscles, located above the knee. They are responsible for the extensibility of the lower leg. At the back of the thigh, the muscles responsible for the hip and knee movement are the hamstrings muscles, locate under the gluteus maximums behind the hip bone and attached to the tibia at the knee. Finally, we have the calf muscles that provide pivotal moment to the movement of the ankle, foot and toe.

Musculoskeletal Levers

When these muscles contract, the force of the muscle contraction acts as lever, maximizing the work done by this force. Levers is any fixed axis capable of rotate around a point and distribute the force acting weight over the lever. In the body joints work as the rotation point, bones as lever and the muscle provides the force. The positioning of the axis point, force and resistance define three type of lever.

A type 2 lever is present in the inferior limbs during movement, as when the body weight is place over the tiptoe of the feet and the muscles of the posterior leg lift contract and lift the calcaneus, causing the resistance to be between the ground force and the axis point.

A type 3 is the most common type of lever in the body and can also be found in the inferior limbs. This type of lever occurs when the force is located between the axis point and the resistance bearing weight, while the force goes upward. This happens in the quadriceps and hamstrings contraction as this allocates between the pivot point (knee joint) and the ground force reaction.

Mechanics during walking

During walking the initial contact is given when the calcaneus contacts the ground at heel-strike, here the ground reaction force (Fr) is posterior to the ankle joint and the knee is extended so the Fr lands anterior to the knee and hip. Next, at foot flat as the knee joint goes into flexion, the Fr is posterior to the ankle and anterior to the hip, but the knee joint becomes posterior reaching the mid stance, where here Fr is analogous to a neutral stance, where the ground reaction is in line with gravity and coincidental in action. The body weight then is transferred over the foot toe, making Fr to move anterior to the ankle, anterior to the knee joint and posterior to hip (Figure 1).

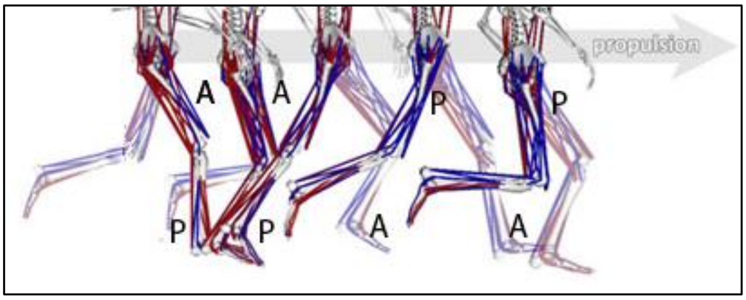

Figure 1: Ground force reaction of the mechanics during walking (font: Hamner et al., 2010 (adapted))

Methods

We developed a simple mathematical model of the musculoskeletal system to examine the inversive dynamics properties of the human movement and other biomechanical problems. The model was based on the lower limb bones (pelvis, femur, tibia, patella and foot), joints (hip, knee and ankle) and the main leg muscles (quadriceps and hamstring). We then analysed two sets of gait data from patients with different walks where we examined the ground reaction force (Fr), the ankle angle, the hip angle, and the knee angle. The first set of gait data was the control, collected from a subject with a normal walking ability. This patient was 1.784 meters in height and 74 kg in weight. The second was from a patient with a walking disability caused by a leg length discrepancy (LLD). The patient was 1.512

meters in height and 64.8 kg in weight. In these data, the joints angles and Fr were collected with the software *Visual 3D* and the model was calculated and treated with MATLAB and Microsoft Excel.

Mathematical Model

The mathematical model (Figure 2) of the musculoskeletal system works based on solving the muscles redundancy problem between the torque caused by the ground force reaction (Fr) and the joints moments and muscle forces, that act in opposite direction from inverse dynamics.

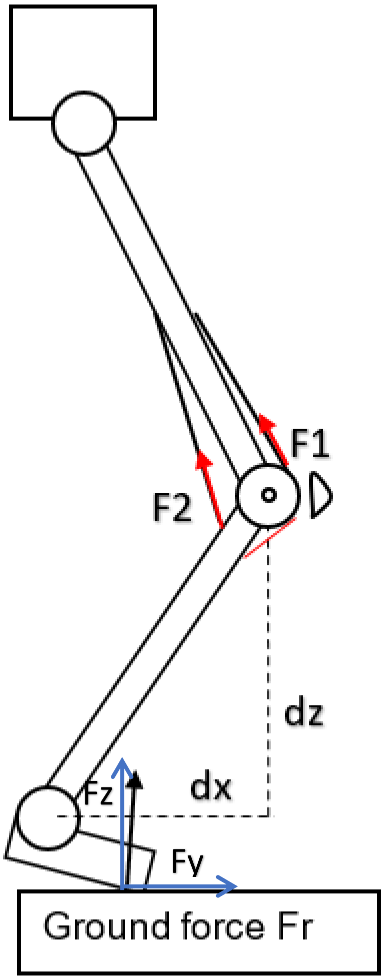

Figure 2: draw of the mathematical model

The force acting on the leg are as follow:

- The upward ground reaction force, acting component that is perpendicular to the pressure surface were feet contacts. (Fr)

- The vector decomposition of the Fr in horizontally and vertically forces. (Fz and Fz)

- The vertical and horizontal moment distance of the torque caused by Fr. (dz and dx)

- The tension or muscle force F1 in the knee extensor muscle, i.e. the quadriceps.

- The tension or muscle force F2 in the knee abductor muscle, i.e. the hamstrings.

Since the static equilibrium condition is assumed here, the summation of the torque of the knee caused by external force Fr and the internal muscle forces F1 and F2 vectors components is zero. This torque caused by the ground reaction can be described by Fr and the moment reaction (Mr). More precisely:

$$T_{\mathrm{fr}} = Fr * Mr \tag{1}$$

However, this vector Fr to be measured it has to be decomposed into a horizontal and vertical vectors and Mr is the perpendicular distance between the Fr point of pressure and the knee angle, with respect to the tibia and vertically to the knee. These horizontal and vertical vectors components distance were calculated by:

$$dy = sin(\theta) * lenght \tag{2}$$

$$dz = cos(\theta) * lenght \tag{3}$$

So for the equation of the torque (T_{fr}) acting at the knee, where $T = M$

$$T_{\mathrm{fr}} = (Fr_{\mathrm{y}} * d_{\mathrm{y}} + Fr_{\mathrm{z}} * d_{\mathrm{z}}) \tag{4}$$

To balance this torque, the main internal muscles (quadriceps and hamstrings) act to maintain the equilibrium using contraction force to pull the bone structure that are attached to acting as a lever.

$$T_{\mathrm{m}} = \boldsymbol{F1} * M1 + \boldsymbol{F2} * M2 \tag{5}$$

Here, the F1 and F2 are unknown muscle forces and M1 and M2 are the Moment, which is the distance of the muscles acting force to the point of rotation. To find this unique solutions F1 and F2 it requires the utilization of optimization techniques that assume that the sum of the muscle forces squared is minimized. The final equilibrium equation is then given by the following expression:

$$\boxed{\text{Equilibrium equation}} \begin{cases} Fr * Mr = \boldsymbol{F1} * d1 + \boldsymbol{F2} * d2 \\ Min(\boldsymbol{F1}^{\mathbf{2}} + \boldsymbol{F2}^{2}) \end{cases} \tag{6}$$

Results

We compared the results obtained from the collected data and optimization of the analytical expression (5) of the gait data of both control and the leg length Discrepancy (LLD). Figure 3 shows the graphs of the angles from the Hip, knee, and ankle of the control and LLD group. Figure 4 is the graphs of the ground forces reactions (Fr), Figures 5 and 6 is the calculated graphs obtained from the mathematical model that shows the muscles force representation of the extensor (F1) and abductor leg muscles (F2).

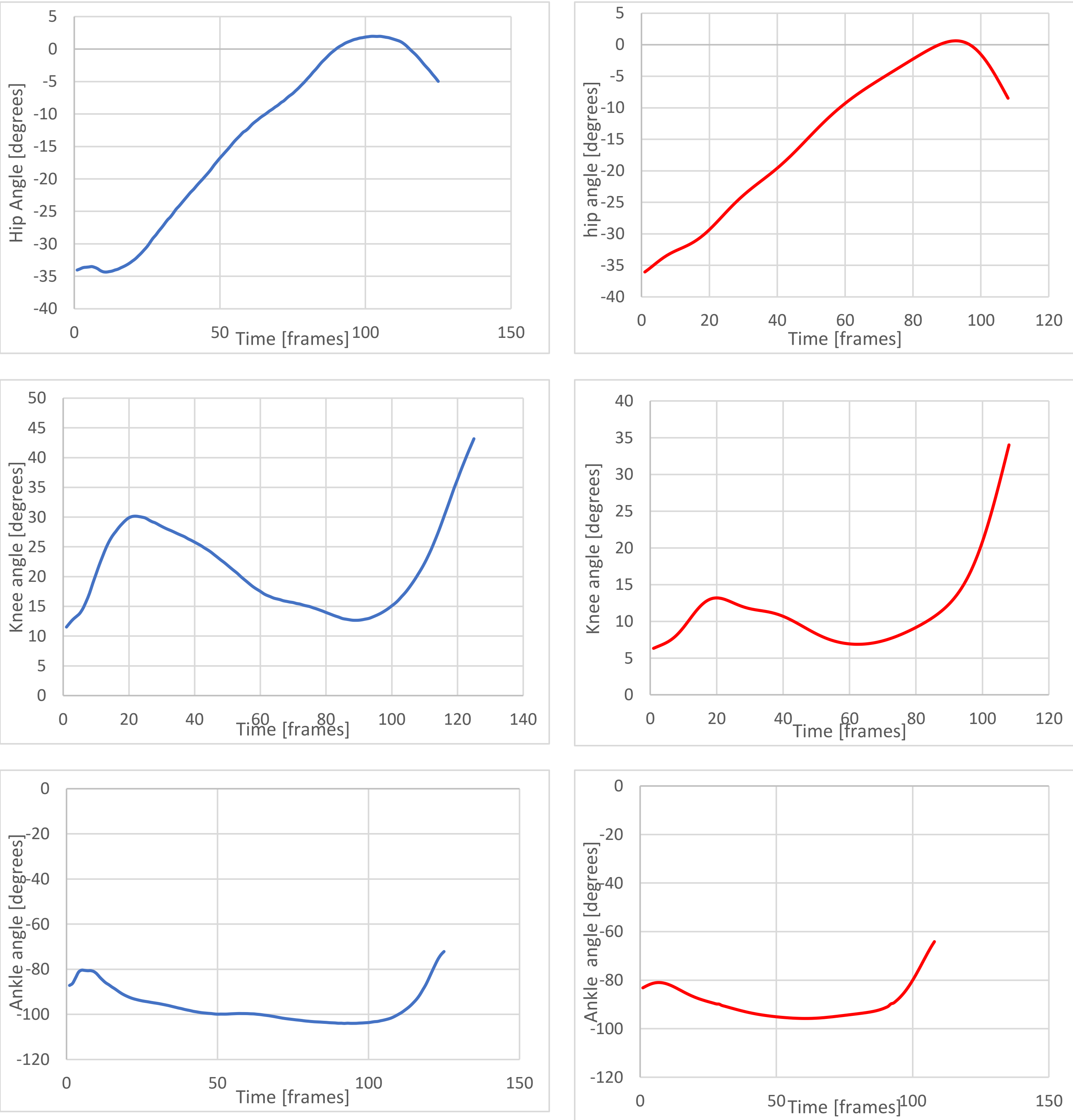

Figure 3: Lower limbs joints angles during the walking of both groups control (left) and LLD (right)

From Figure 3 we see all the joints' angles during both walks of the control and LLD. The functions here are similar in shape, however in the LLD data we can notice tiny differences. In the control data, first the hip angle increases over time, reaching midstance position, then reaching toe-off positioning the angle reaches zero degrees. For the LLD case we can notice a more sudden drop in the hip angle after this maximum extension. For the ankle we see a range of motion around 20 degrees, with an extension during heel strike and an abduction during toe-off followed by a rise during midswim. This, however, is smaller and faster transition on the LLD gait data, indicating flat foot and an earlier toe-off. Finally, the major affected joint was the knee joint, here we can see a rise in the angle as the leg comes into foot flat position, after this the angle decreases to midstance, finally rising again from heel-off to swing until the next heel-strike. This, however, in the LLD leg did not happen. The knee in the first phases is more extended and barely flexes and is followed by premature extension and then flexes only during leg swing. In the next figure we can watch how these joints angles impact the ground force reaction (Fr) in both normal and LLD gait data.

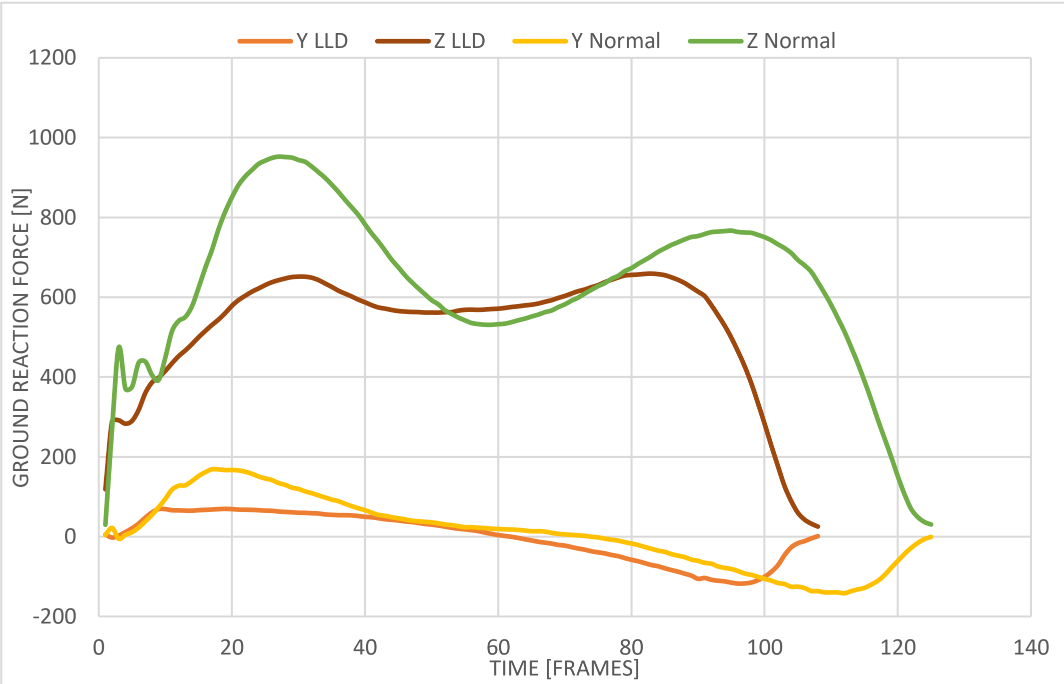

Figure 4: Ground force reaction of the y and z direction of the Normal and LLD gait data

In Figure 4, we can observe that the ground force Fr is divided by the coordinates Y and Z direction. Looking at these ground reactions forces we can conclude that during the normal walking, the force during the first contact with the ground is high, at the heel strike, dropping then significantly during foot flat and midstance, rising again from heel off to toe off in the z direction.

For the Y direction the Fry does a small rise during heel strike to midstance and then the direction is changed to the opposite direction with a similar magnitude from midstance until toe-off. For the LLD data in the Z direction, the Fr is smaller in magnitude and does not have such distinct extremes as the control data and maintains almost constant around the 600N during the walking cycle. However, for the Y direction, the Fry behaves differently, having a minor increase during heel strike but then in the opposite direction having the same magnitude as the Fry of the normal data during the toe-off. The following graphs show the force generated by the internal muscle to contradict the torque caused by these ground force reactions, and its relation to the knee angle joint.

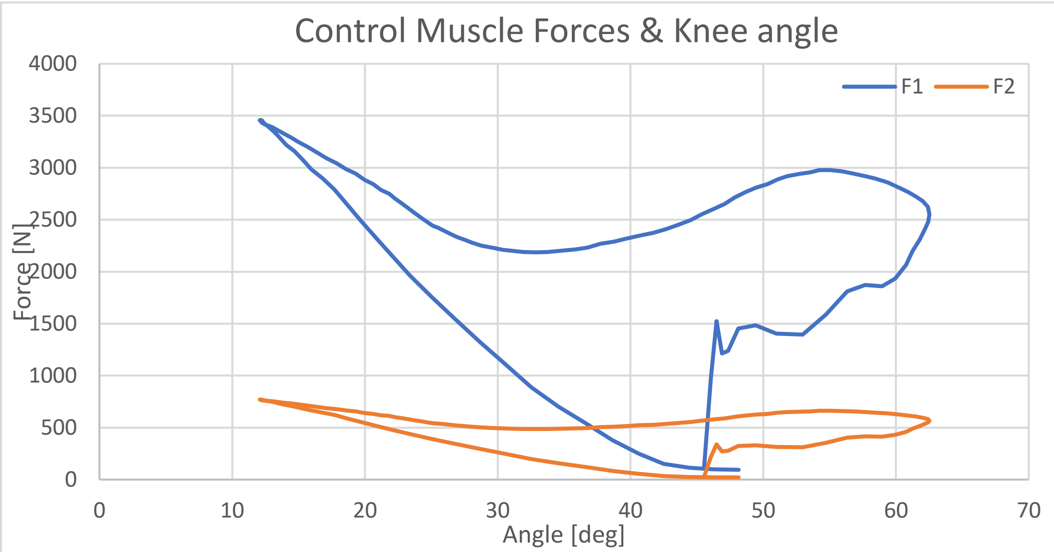

Figure 5: f1 and f2 muscles forces versus knee angle (respect to vertical) in degrees of the control gait data

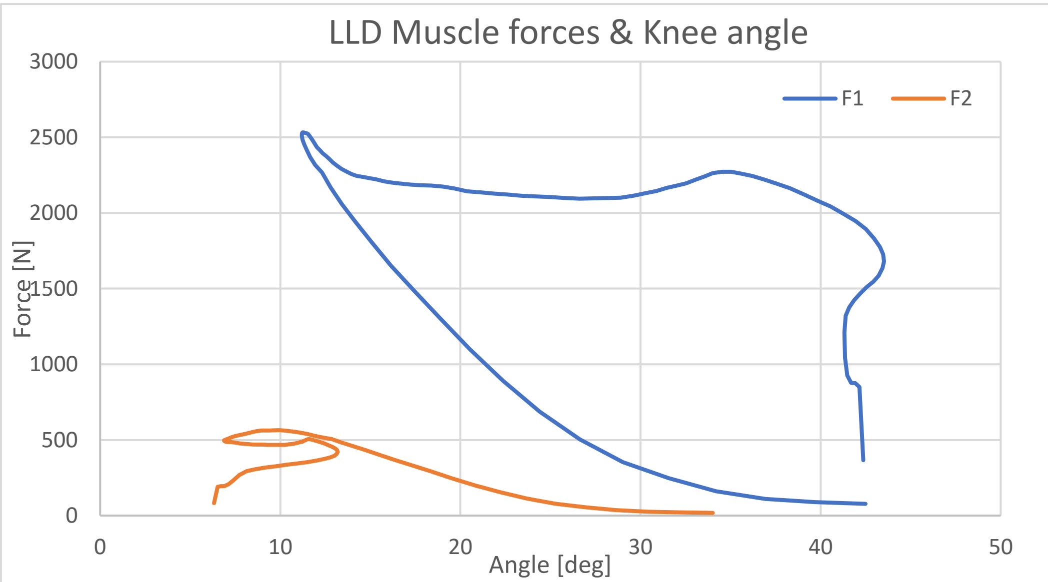

Figure 6: f1 and f2 muscles forces versus knee angle (respect to vertical) in degrees of LLD data gait data

From Figures 5 and 6 we can notice how the group muscles quadriceps and hamstring react to both patient walking cycles. These graphs show us the plot of muscle forces and the knee angle formed between the vertical axis from the knee and the tibia. In Figure 5, the control data, the force F1 starts by increasing rapidly, while the angle decreases between 50 and 10 degrees marked by the heel strike to midstance. Here, the force then decreases as the knee angle starts increasing between heel-off reaching toe off. In this moment then, in combination with F2 the muscles forces project the body forward causing the leg swing and the force and angle drops.

In the LLD data (Figure 6), the forces act similarly, however with less magnitude, less extremes, and a smaller knee angle. F1 increases rapidly during the heel strike moment, but then the force maintains almost constant during midstance to toe-of however. Here, the force of the F2 muscle is not seen.

Discussion and conclusion

This work revealed how the study of the inversive dynamics represents an effective approach to the analyse of human movement and how a leg length discrepancy can have impacts in the dynamics of the joints, in the ground force reactions and in the leg muscles. In addition, we study how the flexion angle of the knee joint can relate to the muscle force in particular the hamstrings.

Our mathematical model showed good results for the understanding of the role of the main abductor and extensor leg muscles. However, there is still room for improvement of this model since the model had several limitations such as incorrect number of muscles, neglect of leg weight, inaccurate size of leg length and pivotal distances. Nonetheless, the calculation of the ground reaction torque was still theoretical correct since it was most based on a real-life data collection.

This real-life gait data collection from the data control (normal) and Leg Length Discrepancy (LLD) also showed us the relation of how the lower musculoskeletal joints behave during movement, and how different the motion of a patient with leg length discrepancy can be from a normal individual. For instance we noticed in the graphs of the joints hip, knee, and ankle (Figure 5), the biggest difference is seen in the knee that during the first phase of walking cycle, does not have a proper flexion. In addition the ankle and hip also presented slight differences, such as a smaller angle of extension for the ankle and a more sudden drop after flexion in the hip angle. We also looked upon the ground force reaction (Fr) that showed to have an equal distribution of force over the feet, with similar highs and almost constant. This leads us to conclude that the LLD patient when it walks, he keeps the leg extended and that during the walking cycle is unable to place the force over extremities of the feet, leaving the feet flat and therefore does not going into toe-off where would propel the body forward. This is why in our model Figure 6 showed a drop in the abductor muscles (F2) in the final phase not utilizing the hamstring muscles for this work.

Our model suggests that during a normal walking cycle, the person uses more the quadriceps and the hamstrings muscles during heel strike and then toe-off, where the hamstring pushes the body forward (Hamner et al., 2010). However, in the LLD gait data, this is not necessarily true and the body ends ups compensating this movement by lifting the leg only utilising the quadriceps and neglecting the hamstrings (Thote et al., 2015). However, when the legs are not the same length, the body compensates, altering the stress across the joints. This can lead to many complications such as knee arthritis as well as shortening of the hamstring muscle, which can limit the ability to stop hip flexion and results in structural asymmetry and future non-structural scoliosis. The recommended treatment for LLD, since the patient is an adult, is placing an orthotic inside the shoes or having shoes for the LLD, that have proper heel lift, so the legs are the same length to correct the discrepancy and improve the gait (Mark A. Caselli & Edward C. Rzonca, 2002).

The goal with this work was to develop a simple mathematical model to examine the inversive dynamic properties of the musculoskeletal system during the human movement, undertake a biomechanical analyse of a postoperative LLD problem and suggest a rehabilitation technique. This was done by creating a mathematical model for the equilibrium equation of the ground force reaction torque and the internal leg muscles. Then, through optimization methods, we achieved the unique solutions for the internal muscle forces (F1 and F2) and then analysed the involvement of these muscles during walking in respect to the knee angle. After, we analysed the difference between the other angles and the components of the Fr force during the normal walking control data and the LLD data, which showed us the risks and the potential future consequences. Although this model has several limitations, discussed above, still proved to be suitable for this study and analyse of data.

In conclusion, the analyse of inverse dynamics is important due to the convenient calculation properties of the human motion, such as the study of muscle forces, moments, and balance of torque. Currently, thanks to modern computational optimization techniques, is possible to study more accurately and faster, the biomechanics involved, thus allowing to advance more during research and provide more suitable rehabilitation solutions.

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