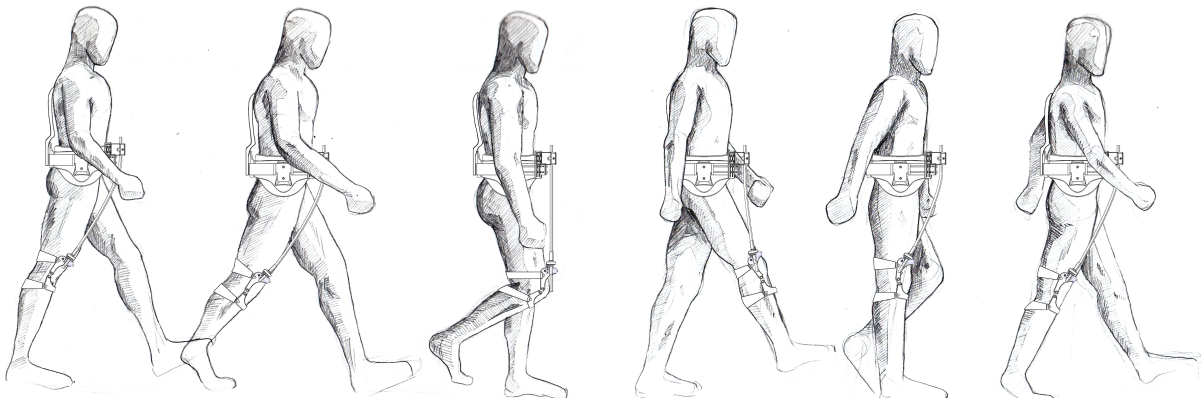


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Design and Evaluation of a Passive Hip Exoskeleton to Reduce the Energy Cost of Human Walking

Design and Evaluation of a Passive Hip Exoskeleton to Reduce the Energy Cost of Human Walking

By

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Writing these sentences means I am almost finished writing my thesis: the ‘finale’ of my Master’s degree and therefore my life as a student. I do not want to use this space to formulate dramatic sentences about my ups and downs during this process, but rather want to thank the people that helped me during my thesis.

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All good things come to an end.

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Abstract

A passive exoskeleton was designed with the goal of reducing metabolic cost of human walking by augmenting moments around the hip joint. A novel approach, using leaf springs parallel to the human leg, was proposed. A kinematic model including man-machine interface compliance was optimized and results showed a maximal reduction of metabolic cost of 60%. A prototype was built and included a pelvic part, vertically fixing the leaf springs, and a thigh part, accommodating four degrees of freedom. The performance of the prototype was experimentally evaluated in treadmill trials. Muscle activity, metabolic energy and whole-body kinematics and kinetics were measured during normal walking, walking with the exoskeleton with and without springs. Walking with the active exoskeleton resulted in a negligible small increase in metabolic energy expenditure compared to the exoskeleton without springs.

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List of Abbreviations

HS	—	Heel Strike
FF	—	Foot Flat
HO	—	Heel Off
TO	—	Toe Off
DOF	—	Degree(s) of Freedom
PA	—	Pelvic Attachment
TA	—	Thigh Attachment
HJC	—	Hip Joint Center
ASIS	—	Anterior Superior Iliac Spine
PSIS	—	Posterior Superior Iliac Spine
PS	—	Pubic Symphysis
ATP	—	AdenosineTriphosphate, the molecular unit of energy transfer
FEA	—	Finite Element Analysis
CAD	—	Computer -aided Design
GPa	—	Gigapascal, common unit for stress-related properties e.g. Young's Modulus
CFRP	—	Carbon Fiber Reinforced Polymer
GFRP	—	Glass Fiber Reinforced Polymer
EMG	—	Electromyography
$\dot{V}_{O_2}$	—	Rate of oxygen consumption
$\dot{V}_{CO_2}$	—	Rate of carbon dioxide consumption
PHE	—	Passive Hip Exoskeleton

Chapter 1

Introduction

Despite many technological inventions such as the bike and the car, walking remains a common means of overland transport worldwide. Benefits of walking compared to vehicles are: no necessity of external power supplies, access to obstacle-ridden environments and a small footprint for agile movements. Moreover, it is free and healthy. Regular walking can reduce the risk of coronary heart disease and strokes [1], improves mental health and aids sleep [2].

Hence, making walking more efficient can be very promising. Humans will be able to cover larger distances on foot with the same amount of effort. This could make it more appealing to cover small distances on foot instead of using the car, reducing carbon emissions [3]. Making walking more efficient can be helpful for other target groups too. Foot soldiers carrying heavy loads encounter severe stress during prolonged marching which can lead to clinical disorders [4]. Compared to young people, elderly have increased muscle weakness [5–6], excessive co-activation of antagonist muscles [7] and a change in neuromuscular control [8], which causes a higher cost of walking [9]. Though, this thesis focuses on regular walking of healthy individuals. When improvements are gained for this target group, the research can be broadened and presumably help the aforementioned groups too.

A popular means for increasing efficiency in walking are active lower-limb exoskeletons. These systems provide torque at certain joints by using actuators to augment the human. Many of these exoskeletons however do not successfully reduce energy costs. The low power-density of current electric motors and the excessive weight of these exoskeletons normally outweigh the support provided to reduce energy expenditure. Therefore it is not surprising that the first exoskeleton to reduce the energy cost of walking was an exoskeleton tethered to a non-portable pneumatic air supply [10]. The first autonomous active exoskeleton achieving a reduction in energy cost made use of an actuated ankle-foot exoskeleton [11]. Other autonomous exoskeletons were only able to gain reductions compared to their unpowered equivalent [12–14,].

The main problem with active exoskeletons is their dependence on battery sources. A finite operating-time limits human walkers to a certain area, as much as a tethered device limits the user to a treadmill. For these reasons research is conducted into passive exoskeletons. However, these unpowered devices require a different strategy. Instead of delivering torque from a power source, energy in the body has to be recycled to improve energetic efficiency. The first passive exoskeleton that managed to reduce the metabolic cost of walking used a lightweight ankle design with a spring acting parallel to the calf's muscle [15]. A biologically inspired exoskeleton

using multiarticular artificial tendons [16] and an exoskeleton with springs in the three lower-extremity joints [17] could not reduce metabolic cost with respect to normal walking. It can be often seen that the increased metabolic cost due to the added mass is still outweighing the metabolic advantages of the exoskeleton. Especially adding mass to the foot bears a relatively high cost on net metabolic rate, an increase of 7.5%-8.5% [18–19] per kilogram of total added mass. This is around 4 times as costly compared to adding mass to the waist, considering an increase of 1.9%-2.1% [19–20] per kilogram of total added mass. Therefore, an exoskeleton augmenting the human hip joint is more likely to decrease the metabolic cost of walking than for example an ankle exoskeleton. Eventually, when successfully applied to the hip joint, focus can be shifted to use the same technique for multiple joints to decrease the overall metabolic cost.

The aim for this research is to develop a passive lightweight hip exoskeleton that minimizes the mechanical work at the hip joint to increase mobility and decrease the metabolic energy expenditure of healthy individuals during level walking. The emphasis is on gaining reductions in the sagittal and frontal plane.

To achieve this, a hip exoskeleton bearing a leaf spring parallel to the upper leg was designed. Optimizations were carried out to find the best setup to achieve metabolic reductions. A prototype was manufactured and subsequently used for gait trial experiments on metabolic cost, kinematics and muscle activity.

1.1 Human Gait

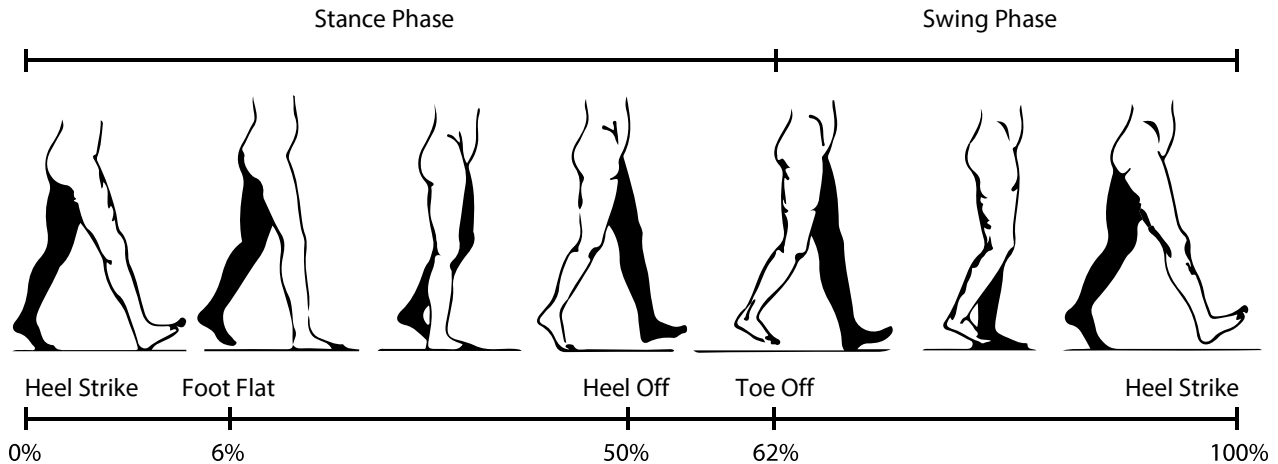


Figure 1.1 — Human walking gait cycle and events, beginning and ending at HS. Percentages are approximated averages.

Figure 1.1 describes the numerous events during one full gait cycle (one stride, two steps) of normal human walking for one leg. It should be noted that the percentages are approximated and can differ among subjects, walking speeds, ages and well-being. One gait cycle is divided between the stance phase (0-62%) and the swing phase (62%-100%), which indicates the

swinging of the leg through the air. The stance phase begins with Heel Strike (HS) and is shortly followed by Foot Flat (FF). In late stance, first the heel is lifted during Heel Off (HO) and later followed by Toe Off (TO) entering the swing phase until the heel makes contact again to begin a new cycle.

1.2 Biological Hip Joint

The human hip joint, or acetabulofemoral joint, is a ball-and-socket joint and has three Degrees of Freedom (DOF). In the sagittal plane, the gluteus maximus and the iliopsoas, among others, are responsible for respectively extension and flexion. In the frontal plane, the gluteus minimus, medius and maximus and the adductor magnus, longus and brevis are responsible for respectively abduction and adduction. Several muscles mentioned before are responsible for rotations in the transverse plane.

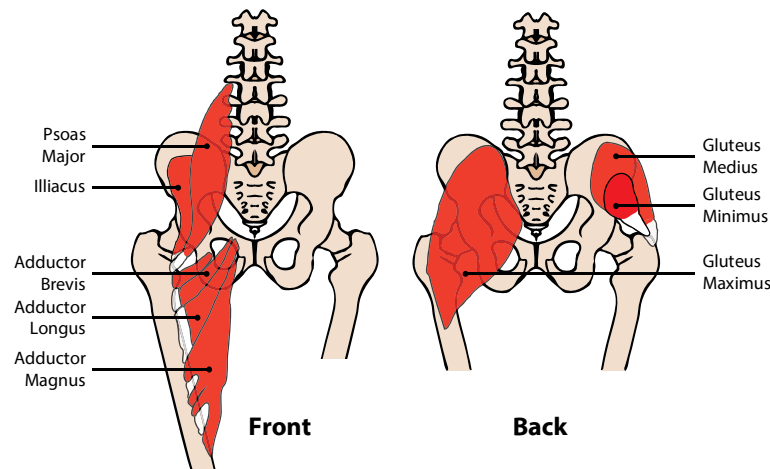


Figure 1.2 — Human muscles surrounding the hip. Psoas major and iliopsoas are key for flexion, gluteus maximus (buttocks) for extension. Abduction is mainly due to the gluteus medius and minimus muscles, adduction due to the adductor muscles.

1.3 Kinematics and Kinetics

Kinematic and kinetic data (subtracted from [21]) of the hip for healthy individuals walking at self-selected speed are shown in Figure 1.3. In the sagittal plane at HS the hip is in a 26 degree flexed position, goes to a -13 degree extended position during stance as it returns to the initial position during swing. Whereas in the frontal plane, the initial posture is around zero degree after which the hip first abducts during stance and finally adducts in swing.

By applying inverse dynamics three-dimensional kinematic data and ground reaction forces can be post-processed into joint torques and powers. In the sagittal plane extensor muscles are active in the first half of stance and flexors in the second half.

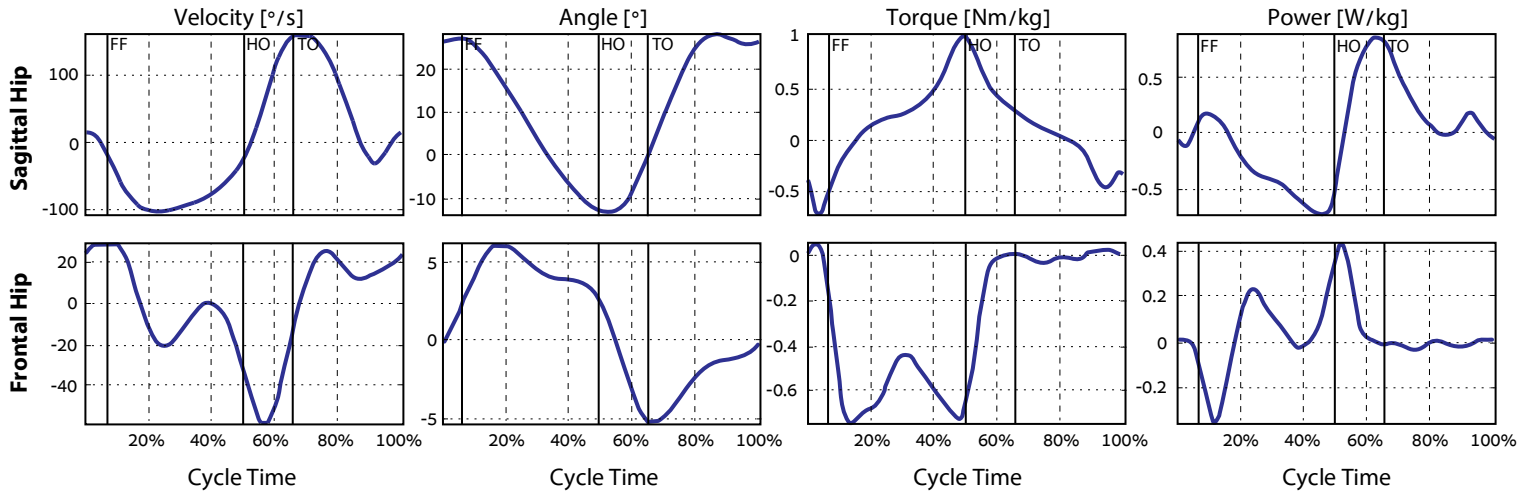


Figure 1.3 — Human gait kinematics at self-selected speed. Torques and powers are normalized to body mass; the time is shown as percentages of gait cycle time; Sagittal and Frontal indicate human anatomical planes. Data subtracted from [21].

During this first half, the extensors stabilize the posture of the trunk by preventing it from flexing forward after HS. The flexor moment during the second half has two functions: stabilizing the trunk and to achieve propulsion of the thigh into swing [22]. This latter half of flexion results in a period in which negative work is being done by the affecting muscles as the joint velocity is negative (extending) but the joint torque positive. In the frontal plane, only abduction torques are being observed which occur during stance. These torques are induced by the gravitational forces of the upper body that cause an adducting moment.

Chapter 2

Methods

2.1 Working principle

Human beings are skilled walkers, they have been slowly optimizing their musculoskeletal and neurological systems for this particular task from when they were 2 years old and take around 10.000 steps per day [23]. They choose step length, frequency and width exactly as to minimize their net metabolic rate [24–25]. Whereas choosing a higher stride rate increases metabolic cost due to an increase in mechanical work, and a lower stride rate due to a decrease in mechanical efficiency of the muscles. This makes it difficult to improve on the efficiency of locomotion by adding a mechanical system.

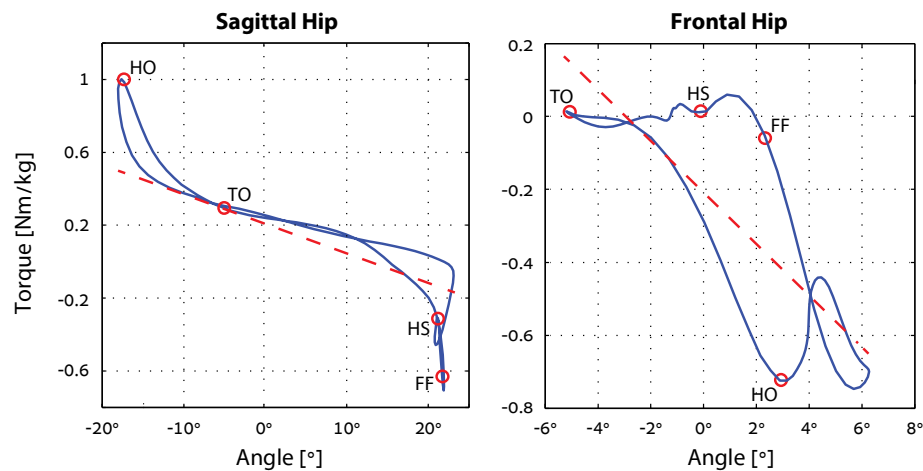


Figure 2.1 — Characteristic torque-angle plots and fitted linear springs, for the sagittal (**left**) and frontal (**right**) hip planes for able-bodied walking at self-selected speed (blue solid) and linear springs (red dashed). Data subtracted from [21].

In theory, no net mechanical power is needed for level walking at steady speed, as kinetic and potential energy supplement each other during each stride [26]. This is shown in simulation models of purely passive spring-loaded legs, where the springs store and return energy during gait and no net change in energy is shown [27]. However, humans do expend metabolic energy during walking: to restore energy that has been dissipated in passive motions of soft tissue and because of the fact that negative work performed by muscles also requires metabolic energy. Negative work occurs when torque and velocity have opposing directions, a ‘braking’ motion, for example when the swinging leg has to be redirected during gait. Mechanical springs do not have this problem because they can absorb and return energy passively. By letting mechanical springs absorb energy during gait-phases where negative work is done and reuse this energy when

positive work is needed, metabolic profits can be gained. An interesting study showed that it is possible to generate electricity from level-ground walking by means of activating a generative energy harvester during negative work of the knee [28]. Instead of harvesting this energy, one could imagine that this energy could be stored and directly used to augment the positive power in joints of the lower-limbs. In the hip, negative work occurs during stance and late stance in the sagittal and frontal plane respectively, according the negative power profiles (Figure 1.3).

Wang et al. [21] investigated the use of parallel springs to lower the power and torque requirements of actuators for exoskeleton purposes. This resulted in a reduction of the root mean square power of 50% and 45% for hip flex/extension and hip ab/adduction respectively. Although this research was focusing on exoskeleton actuators, this principle can be applied to augment human joints as well. In Figure 2.1 it can be seen that torque-angle characteristics of the hip joint, especially for the sagittal plane but also in the frontal plane, for able-bodied walking at self-selected speed have a very appropriate shape for applying a (non)linear spring.

2.2 Design approach

In order to minimize the mechanical work at the hip joint, a mechanism has to be introduced to alleviate the physiological burden at this joint. Springs are passive elements with good properties for efficient energy storage and therefore a logical choice.

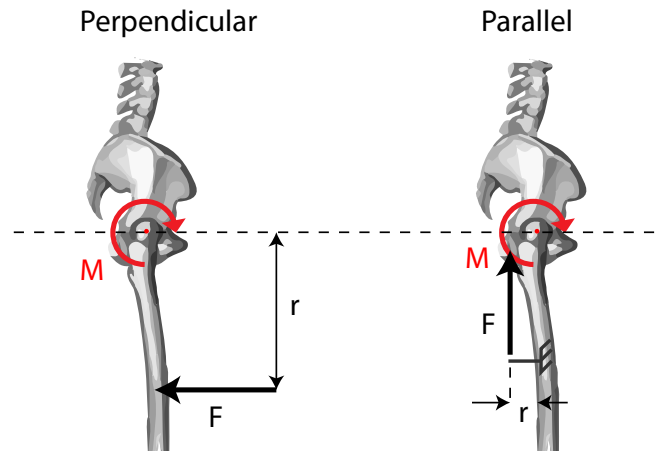


Figure 2.2 — Two ways to apply external moments to the hip joint: creating forces (F) perpendicular and parallel to the upper leg inducing moments (M) around the hip. Moment arm depicted as r .

Two ways to apply external torques to the hip joint in the sagittal plane can be seen in Figure 2.2: applying forces perpendicular or parallel to the upper leg. If compact solutions are desirable, forces for the latter need be big, since the moment arm is small. Hence, perpendicular forces are chosen as a starting point for this study.

One way to exert perpendicular forces is to position a leaf spring parallel to the human upper leg. Suppose this leaf spring is a rod with circular cross-section, connected at the Pelvic Attachment (PA) and Thigh Attachment (TA) by braces (Figure 2.3). At the pelvis the leaf

spring is clamped and therefore has zero DOFs. At the thigh the leaf spring has four DOF, one translational DOF along the longitudinal axis of the leaf spring and all three rotational DOFs. This last property ensures that there are no unwanted moments at the TA and that the moments that are present are pure bending moments of the leaf spring.

2.3 Model Derivation

The process to calculate the residual torques around the hip is shown in Figure 2.3 and can be described as follows:

1. First of all, the human exerts torques τ_G around the hip joint in order to propel itself during gait.
2. This causes the leg to rotate around the hip joint creating angles in 3 planes $\theta_S, \theta_F, \theta_T$. These angles are translated into coordinates for the TA (subsection 2.3.2 Angles to Coordinates).
3. With known coordinates of PA and TA the deflection δ and length L of the leaf spring can be obtained.
4. These values are subsequently used to calculate the forces P in the leaf spring following Bisshopp's theorem (subsection 2.3.3 Bisshopp's Large Deflection Beam Theory)
5. Finally, by multiplying these forces with the moment arm r the torques τ_{Exo} following from the passive exoskeleton are obtained.

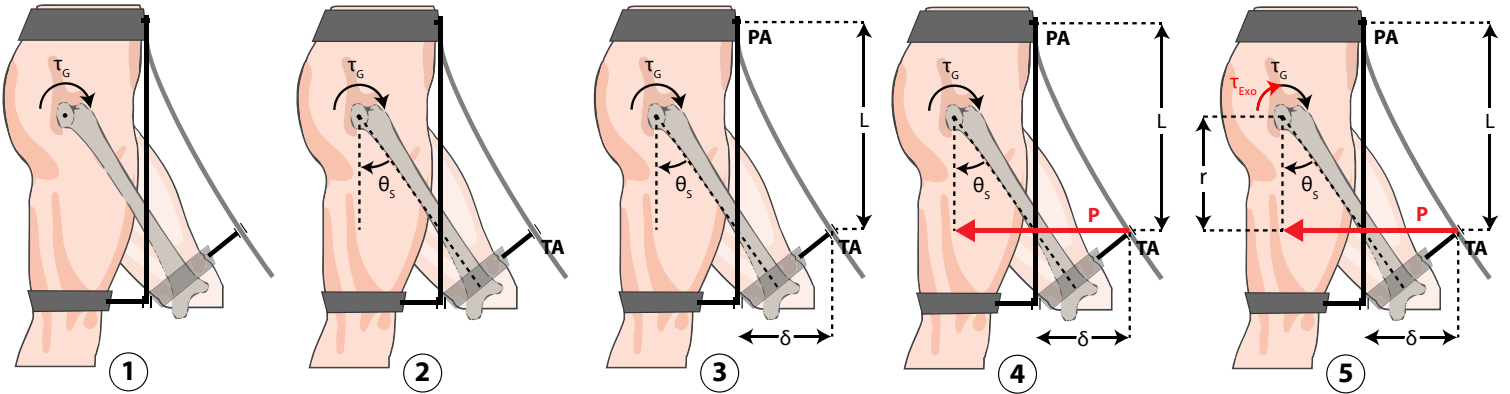


Figure 2.3 — Range of pictures showing the process of calculating the residual torques. (1) Human exerts torques τ_G during walking. (2) The hip angles $\theta_S, \theta_F, \theta_T$ are translated into coordinates for TA. (3) With known coordinates PA and TA, deflection δ and length L of the leaf spring are obtained. (4) These values are used to calculate the forces P in the spring following Bisshopp's theorem. (5) By multiplying with the moment arm r torques following from the exoskeleton τ_{Exo} are calculated.

These values can subsequently be subtracted with the gait torques τ_G to gain the residual torques at time t :

$$\tau_{Res,i}(t) = \tau_{G,i}(t) - \tau_{Exo,i}(t) \quad (2.1)$$

Whereas i denotes either sagittal or frontal plane. Now residual mechanical joint power can be calculated:

$$P_{Res,i}(t) = \tau_{Res,i}(t) \cdot \omega_{G,i}(t) \quad (2.2)$$

Where ω_G stands for the angular velocity of the hip in plane i at time t during gait.

Next, absolute residual mechanical work for one full gait cycle is calculated by taking the time integral over one stride:

$$W_{Mech,i} = \int_0^T |P_{Res,i}(t)| dt \quad (2.3)$$

Where T is stride time.

In subsection 2.3.4 it is explained that not the total mechanical energy is optimized for, but instead mechanical energy is converted to metabolic energy and successively optimized for.

2.3.1 Kinematic Model

A model with seven variables was introduced and can be seen in Figure 2.4. Variables d_1 , d_2 and d_3 are presented to position the two attachment points, PA and TA, with respect to the Hip Joint Center (HJC). Whereas d_1 describes the distance of the PA in positive y-direction from the HJC, d_2 describes the distance of the PA from the HJC in transverse plane and the distance of the TA with respect to the HJC in negative y-direction is denoted with d_3 . Variable r_1 indicates the radius of the leaf spring with circular cross-section.

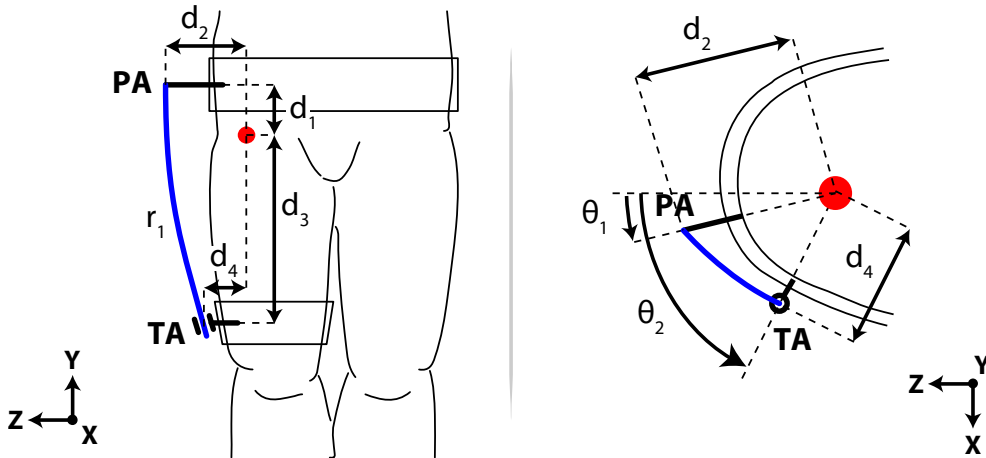


Figure 2.4 — Description of variables for optimization. Variable d_1 describes the distance of the PA in positive y-direction from the HJC (red dot). Variable d_2 together with angle θ_1 describes the distance of the PA from the HJC in the transverse plane. The distance of the TA with respect to the HJC in negative y-direction is denoted with d_3 . The radius of the leaf spring with circular cross-section is described by r_1 . Variables d_4 and θ_2 describe the distance of the TA from the HJC in the transverse plane.

When the body is at rest (zero degrees of flex/extension and ab/adduction) flexion and abduction torques are exerted by the body (Figure 2.1). This means that the leaf spring should initially bent inward in both sagittal and frontal planes to create initial flexion and abduction torques. To achieve this, extra variables have to be introduced. First of all, a new variable d_4 is composed, which represents the distance from the TA to the femur. To create initial flexion and abduction torques d_4 must be smaller than d_2 , which creates a constraint C_1 :

$$C_1: d_4 < d_2 \quad (2.4)$$

To negotiate between these initial abduction and flexion torques, two variables θ_1 and θ_2 are introduced to vary both angles of PA and TA with respect to its lateral position.

2.3.2 Angles to Coordinates

To calculate the coordinates of the TA, angular values in sagittal, frontal and transverse planes should be processed in the right sequence. Conventional, Cardan angles are used to describe flexion-extension, adduction-abduction and internal-external rotation [29–31]. The hip joint, assumed to be a ball-and-socket joint with three DOFs, can be modelled as a chain of revolute joints with link lengths zero as depicted in Figure 2.5. In this representation, the first rotation θ_S resembles flex/extension, the second θ_F ab/adduction and the third θ_T internal/external rotation and therefore is in agreement with the Grood-Suntay convention [31].

Next, Denavit-Hartenberg parameters [32] are presented in Table 2.1 to describe the properties of the different links shown in Figure 2.5. As mentioned before, the first two links connecting the three rotational DOFs of the hip joint, have zero lengths a and e because they are self-aligning to form a proper ball-and-socket joint. Parameters $d_1, d_2, d_3, d_4, \theta_1$ and θ_2 are described in subsection 2.3.1.

Table 2.1 — Denavit-Hartenberg parameters for set-up depicted in Figure 2. Parameter a_i is distance along x_i from O_i to the intersection of the x_i and z_{i-1} axes. Parameter e_i is the distance along z_{i-1} from O_{i-1} to the intersection of the x_i and z_{i-1} axes. Parameter α (alpha) is the angle between z_{i-1} and z_i measured about x_i . Parameter θ_i is the angle between x_{i-1} and x_i measured about z_{i-1} [32]. Angles $\theta_S, \theta_F, \theta_T$ are angles of the human hip joint in sagittal, frontal and transverse plane respectively.

Link	a	e	α	θ
1	0	0	-90°	$-90^\circ + \theta_S$
2	0	0	-90°	$90^\circ + \theta_F$
3	$-d_4$	$-d_3$	180°	$\theta_2 + \theta_T$

The origin of the global reference frame is coinciding at the HJC. Where the **Z**-axis is determined by the line parallel to a line connecting the right and left Anterior Superior Iliac Spine (ASIS), pointing to the right. The **X**-axis defined by the line parallel to a line lying in the plane defined by the two ASISs and the midpoint of the two Posterior Superior Iliac Spines

(PSIS), orthogonal to the Z -axis, pointing anteriorly. Finally the Y -axis is the line perpendicular to both X and Z , pointing cranially [29].

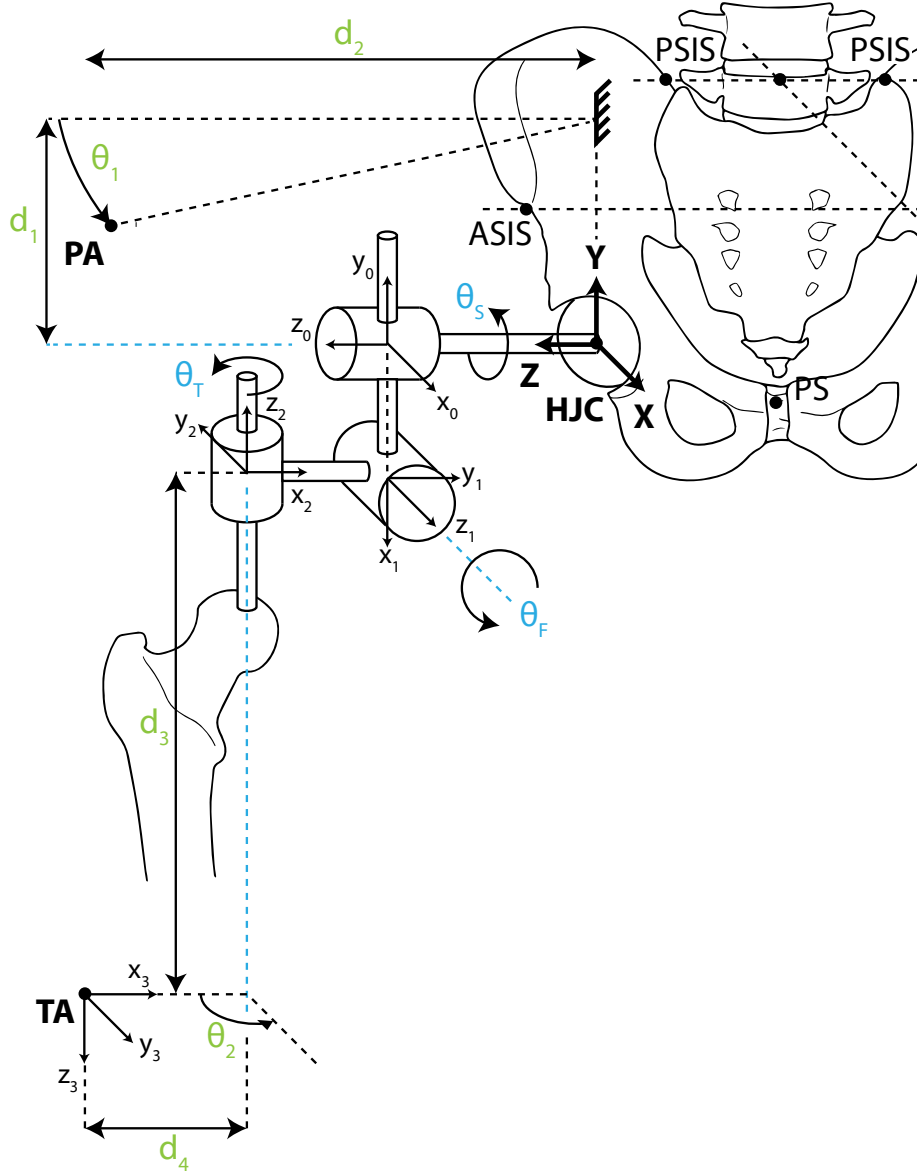


Figure 2.5 — Coordinates of the TA expressed as a kinematic chain. Global reference frame X, Y, Z is coinciding at the HJC. The hip joint is represented as three revolute joints with link lengths zero. First revolute joint has a Local Coordinate System (LCS) x_0, y_0, z_0 and rotates about θ_s (flex/extension). Second revolute joint has a LCS x_1, y_1, z_1 and rotates about θ_F (ab/adduction). Third revolute joint has a LCS x_2, y_2, z_2 with link length d_3 and rotates about θ_T (internal/external rotation). The end effector TA with local cs x_3, y_3, z_3 has link distance d_4 and angle θ_2 . PA is positioned with parameters d_1, d_2 and angle θ_1 .

Now the coordinates can be calculated, starting with the HJC, which is always defined as the origin:

$$HJC = [0 \ 0 \ 0] \quad (2.5)$$

Next, with the presented Denavit-Hartenberg parameters, rotation matrices can be generated in order to find coordinates for TA

$$A_i = \begin{bmatrix} \cos \theta_i & -\sin \theta_i \cos \alpha_i & \sin \theta_i \sin \alpha_i & a_i \cos \theta_i \\ \sin \theta_i & \cos \theta_i \cos \alpha_i & -\cos \theta_i \sin \alpha_i & a_i \sin \theta_i \\ 0 & \sin \alpha_i & \cos \alpha_i & e_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (2.6)$$

where i corresponds to a particular link. Multiplying these rotational matrices together yields

$$T_3 = A_1 \cdot A_2 \cdot A_3 \quad (2.7)$$

which is the rotation matrix in order to calculate the coordinates of TA with respect to the global reference frame **XYZ**, now TA can be calculated as follows

$$TA = T_3 \cdot HJC \quad (2.8)$$

The coordinates for the PA are mainly given by the parameters and can be described as

$$PA = [\sin(\theta_1) \cdot d_2 \quad d_1 \quad \cos(\theta_1) \cdot d_2] \quad (2.9)$$

2.3.3 Bisshopp's Large Deflection Beam Theory

Euler-Bernoulli's classical beam theory can be used to determine the bending of a beam. As can be seen in the top part in Figure 2.6, a beam with a certain length L will have a vertical deflection δ due to a point load $\mathbf{P}$. The Euler-Bernoulli equation describes the relationship between a distributed load $q(x)$ and the beam's deflection:

$$\frac{d^2}{dx^2} \left(EI \frac{d^2 \delta}{dx^2} \right) = q(x) \quad (2.10)$$

However, this original Euler-Bernoulli equation is only applicable to beams with infinitesimal strains and small rotations. In the proposed concept, large rotations and deflections are expected and therefore errors will be disclosed when applying this theory. In fact, for large finite loads, the classical beam theory gives deflections greater than the length of the beam, which is impossible.

A modification of the classical beam theory was presented by Bisshopp and Drucker in 1945 [33] and is explained in Appendix 1: Bisshopp-Drucker Beam Theory. This new theory can be used for large deflections of cantilever beams and takes into account the shortening of the moment arm, see Figure 2.6-bottom. It is assumed that the bending does not change the length of the beam (inextensible) and the Euler-Bernoulli assumption, which states that the curvature is proportional to the bending moment is still applied. Bisshopp and Drucker state that the curvature of the elastic line can be expressed in terms of arc length s and slope angle θ .

$$\frac{d\theta}{ds} = \frac{P}{EI} (L - x - \Delta) \quad (2.11)$$

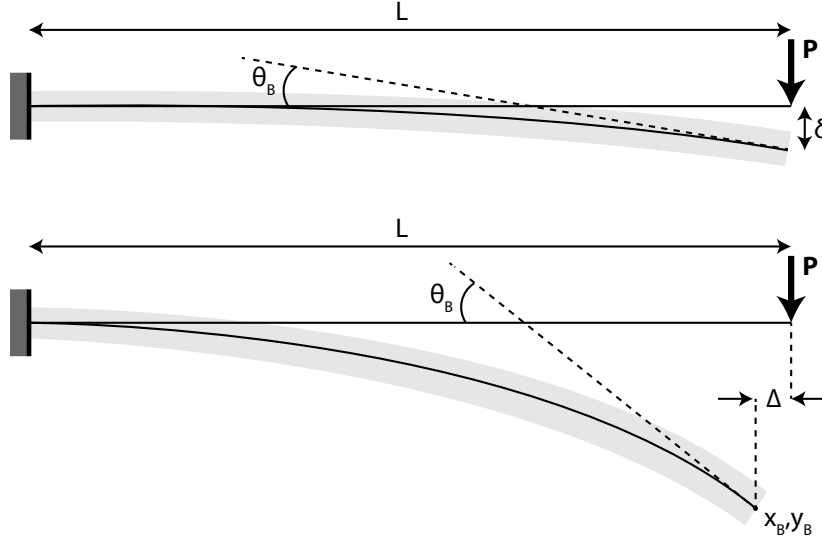


Figure 2.6 — Euler versus Bisshopp, linear versus nonlinear. Euler-Bernoulli beam theory (**top**) and Bisshopp-Drucker nonlinear beam theory (**bottom**). The clamped beam has length L with a vertical force P at the beam end, causing a deflection δ and slope of the bended beam θ_B . The Bisshopp-Drucker theory incorporates an extra term Δ : the horizontal component of the displacement of the loaded end of the beam.

With boundary conditions being zero beam rotation at the clamped site $\theta(0) = 0$ and no curvature at the loaded end $\frac{d\theta}{ds}(L) = 0$. By manipulating equation (2.11) and applying the boundary conditions the force P can be calculated. Subsequently, the moment around the hip caused by the exoskeleton can be calculated (see Appendix 1: Bisshopp-Drucker Beam Theory).

$$\tau_{Exo} = P \cdot r \quad (2.12)$$

Looking back at equation (2.1), component $\tau_{Exo,i}(t)$ can now be described by equation (2.12) and will be referred to as ‘the rigid kinematic model’ for the remainder of this thesis.

2.3.4 Conversion to Metabolic Energy

Muscles have a certain efficiency in which metabolic energy is converted into mechanical work to drive the muscles. To gain insight in metabolic energy expenditure, efficiencies with which muscles turn metabolic energy (ATP) into muscular work need to be derived.

Although stride rate affects the mechanical efficiency of muscles [24], it is commonly believed that concentric muscle actions have an efficiency of approximately 0.25 [34–40]:

$$\eta_+ = 25\% \quad (2.13)$$

This means that 4 Joules of metabolic energy is converted to 1 Joule mechanical work by doing positive muscular work.

Fewer studies are conducted in order to investigate the efficiency of eccentric muscular work or negative work and the ones that are conducted report larger discrepancies. Margaria [34]

considered walking downhill as doing negative mechanical work and the ratio of mechanical work performed to the energy expenditure was about -1.2. This means that -1.2 Joules of mechanical muscle work only requires 1 Joule of metabolic energy. Ryschon et al. [35] used a dynamometer to measure muscle work and monitored ATP synthesis *in-vivo* by investigating P-nuclear magnetic resonance data for one single muscle. The resulting mechanochemical efficiency for eccentric actions was 0.347. Abbott et al. [41] coupled the chains of two ergometers and let two subjects perform as much positive as negative work by letting one subject ride the ergometer and letting the other cyclist resist his movements and oxygen consumption was measured. They found that the ratio of oxygen consumption between positive and negative work is 2.4 for slow pedaling. This means that, assuming a positive mechanical work efficiency of 0.25, the efficiency for negative muscular work is $2.4 \cdot 0.25 = 0.6$. Benedict and Parmenter [39] studied the energy metabolism of women while ascending and descending stairs, they found that the total energy expended while descending stairs (negative work) was about 33% of that expended in ascent. Again assuming a positive mechanical work efficiency of 0.25 a negative mechanical work efficiency of $0.25/0.33 = 0.76$ can be assumed.

Ryschon et al. [35] investigated single muscle ergonomics and used a novel mechanochemical approach; it's hard to compare their results with the other studies mentioned. In the present study the focus lies mainly on full gait with a particular interest on the hip joint which is driven by around 17 muscles [42] that can mutually exchange energy. For these reasons the values above for negative muscular work efficiencies are averaged and results from [35] are left out. Thus, in the present study, the negative muscular efficiency is described as:

$$\eta_- = \frac{1.20 + 0.6 + 0.76}{3} \cdot 100\% = 85\% \quad (2.14)$$

To gain metabolic energy expenditure, positive and negative mechanical work as calculated in equation (2.3) is multiplied with the muscular efficiency for concentric and eccentric muscle action respectively:

$$E_{Meta,i} = \begin{cases} W_{Mech,i} \cdot \eta_-, & \text{if } W_{Mech,i} < 0 \\ W_{Mech,i} \cdot \eta_+, & \text{if } W_{Mech,i} \geq 0 \end{cases} \quad (2.15)$$

Umberger et al. [24] calculated the sum of positive (0.72 [W/kg]) and negative (-0.37 W/kg) mechanical power of hip, knee and ankle during walking trials at 1.3 m/s. Using equation (2.15) this adds up to 3.32 [W/kg] metabolic energy and therefore differs 7% with their reported value for metabolic energy expenditure of 3.10 [W/kg]. Therefore it can be assumed that the approach described is a good representation of mechanical joint work conversion to metabolic energy.

The sum of the metabolic energy for the two planes, sagittal and frontal, were summed to gain the total metabolic energy during one full stride:

$$E_{Meta} = \sum_{i=1}^n E_{Meta,i} \quad (2.16)$$

2.3.5 Prototype and Human Compliance

Up until now, a rigid model is used which means that forces imposed by the Passive Hip Exoskeleton (PHE) are directly transferred to the rigid parts of the human body, as can be seen in Figure 2.7 (left). However, by applying loads to the human body, the soft tissue of the subject will deform. Moreover, the exoskeleton itself contains some compliance as well. Assuming a rigid model can give inaccurate results and many scientific contributions point out that accounting for this effect of compliance should be carefully considered [16,43-44]. Thus, composing a more realistic, compliant model should minimize errors between the theoretical analysis and the experimental results.

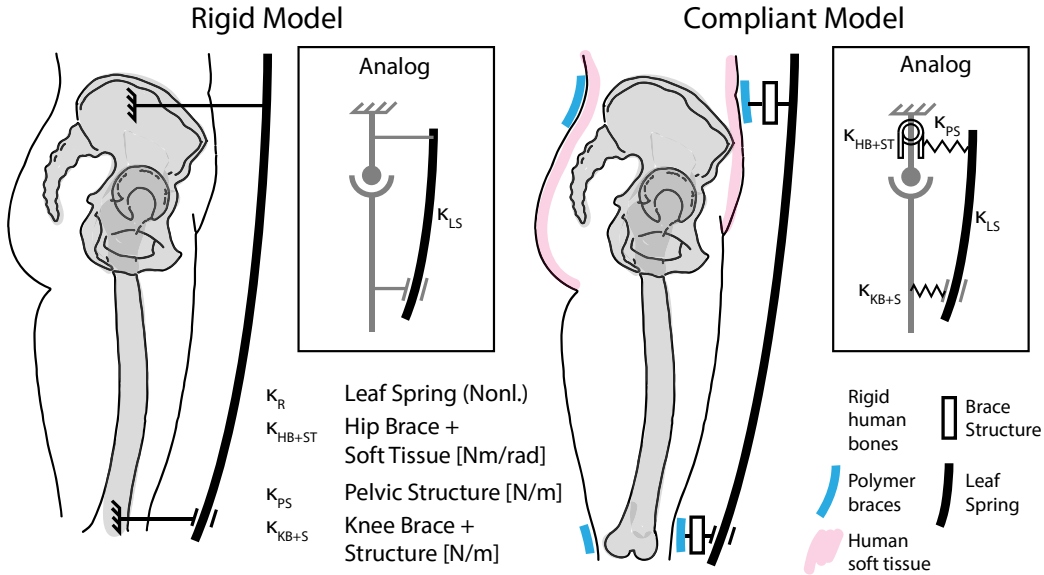


Figure 2.7 — Rigid versus compliant model. In the rigid model (left) forces from the PHE are rigidly transferred to the pelvis and femur. More realistically, in the compliant model (right), transfer of forces goes via soft tissue at the hip (pink), polymer braces (blue) and steel structures (white) at the hip and knee. The rigid model is analogous to a clamped nonlinear leaf spring κ_R ; the compliant model also has a rotational spring at the hip (pelvic brace plus soft tissue: κ_{PB+S}) in series with a linear spring (pelvic structures: κ_{PS}) and a linear spring at the thigh (knee brace plus structure: κ_{TB+TP}).

Maathuis [45] performed experimental tests on the same hip brace (Orthomerica, Newport 4 Hip Orthosis) used for this research. Deformations were measured while being subjected to loads, therefore incorporates compliance of both brace and soft tissue surrounding the hip. Protocol existed of three subjects (82.7 ± 8.6 [kg]) walking at 1.25 m/s. Forces, parallel to the leg where exerted on the brace (Figure 2.8, left), and deformations were measured in terms of degrees. Maximum deflection occurred around 50% of stride and was around 3° and 17° for the no-load and load condition respectively. Associating force exerted was around 400 [N] with a moment arm of 0.16 [m] [45]. Hence, rotational spring stiffness κ_{HB+ST} can be calculated as:

$$\kappa_{HB+ST} = \frac{400 \cdot 0.16}{(17^\circ - 3^\circ) \cdot \left(\frac{\pi}{180}\right)} = 262 \left[\frac{Nm}{rad}\right] \quad (2.17)$$

This value for rotational spring stiffness will be used in both sagittal and frontal planes. Unfortunately, rotational values in the transverse plane were not obtained by Maathuis and therefore are neglected in this model.

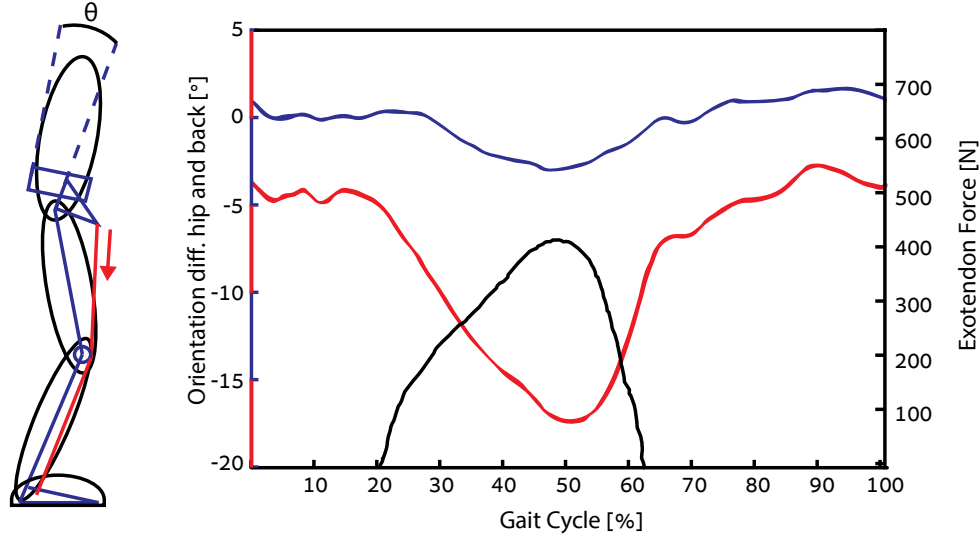


Figure 2.8 — Experimental values obtained for stiffnesses of brace and human soft tissue. **Left:** Representation of experimental setup, with forces causing rotational deformation of the hip brace. **Right:** Results with on the left axis the deformation in degrees between zero-load (blue) and load (red) conditions, right axis forces (black) exerted on the hip brace. Adapted from Maathuis [45].

Next, linear spring stiffnesses of the ‘pelvic structure’ κ_{PS} and ‘knee brace + structure’ κ_{KB+S} were found with Finite Element Analyses (FEA). Static analyses in CAD-software (SolidWorks Simulation, Concord, MA) are conducted in both sagittal and frontal direction with forces and moments following the Free Body Diagram as in Figure 2.9. An example of such a calculation is shown in Figure 2.10, where the pelvic structure is tested with a maximum sagittal load during gait of 56.1 [N] and a moment of 33.66 [Nm] around *Axis 1*. Displacement at the location of loading was -0.516 [mm] therefore the spring stiffness κ_{PS} in sagittal direction is:

$$\kappa_{PS}(\text{sag}) = \frac{56.1}{0.516E^{-3}} = 108.7E^3 \left[\frac{N}{m} \right] \quad (2.18)$$

Consequently, residual spring stiffnesses are:

$$\kappa_{PS}(\text{fro}) = \frac{26.3}{0.418E^{-3}} = 62.9E^3 \left[\frac{N}{m} \right] \quad (2.19)$$

$$\kappa_{KB+S}(\text{sag}) = \frac{56.1}{0.725E^{-3}} = 77.4E^3 \left[\frac{N}{m} \right] \quad (2.20)$$

$$\kappa_{KB+S}(\text{fro}) = \frac{26.3}{0.398E^{-3}} = 66.1E^3 \left[\frac{N}{m} \right] \quad (2.21)$$

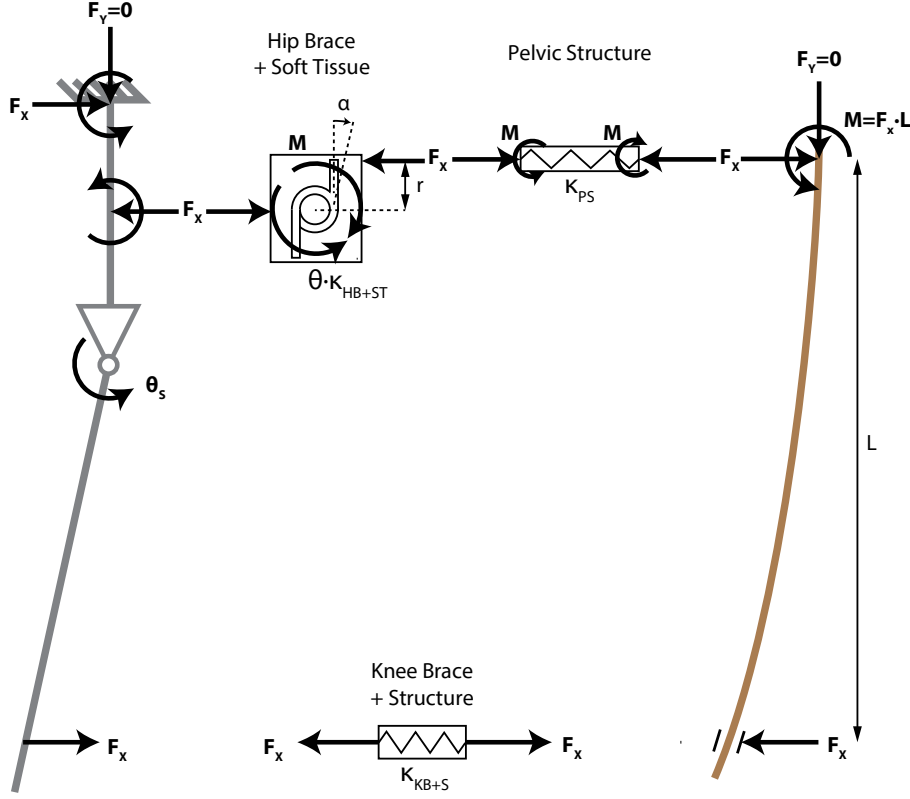


Figure 2.9 — Two dimensional Free Body Diagram of the PHE (sagittal plane). F_x follows from the bended leaf spring, τ_{Sag} is the moment exerted by the human. Grey elements are rigid/fixed elements.

Next, these values are translated into a three dimensional environment in order to cope with the three dimensions created by the human leg swinging in three planes. The compliant series elements at the pelvic region are merged into one equivalent spring. This means the rotational spring κ_{PB+ST} [Nm/rad] has to be approximated as a linear spring:

$$\alpha = \frac{(F_x \cdot L) + (-F_x \cdot r)}{\kappa_{HB+ST}} \quad (2.22)$$

$$\kappa_{HB+ST}(linear, sag) = \kappa_{HB+ST}(linear, fro) = \frac{F_x}{\tan(\alpha) \cdot r} \left[\frac{N}{m} \right] \quad (2.23)$$

where the parameters described in (2.22) and (2.23) are according Figure 2.9.

Equivalent spring stiffnesses can be composed for both sagittal and frontal plane, as described in Figure 2.11:

$$\frac{1}{\kappa_{sag}} = \frac{1}{\kappa_{HB+S}(linear, sag)} + \frac{1}{\kappa_{PS}(sag)} \quad (2.24)$$

$$\frac{1}{\kappa_{fro}} = \frac{1}{\kappa_{HB+ST}(linear, fro)} + \frac{1}{\kappa_{PS}(fro)} \quad (2.25)$$

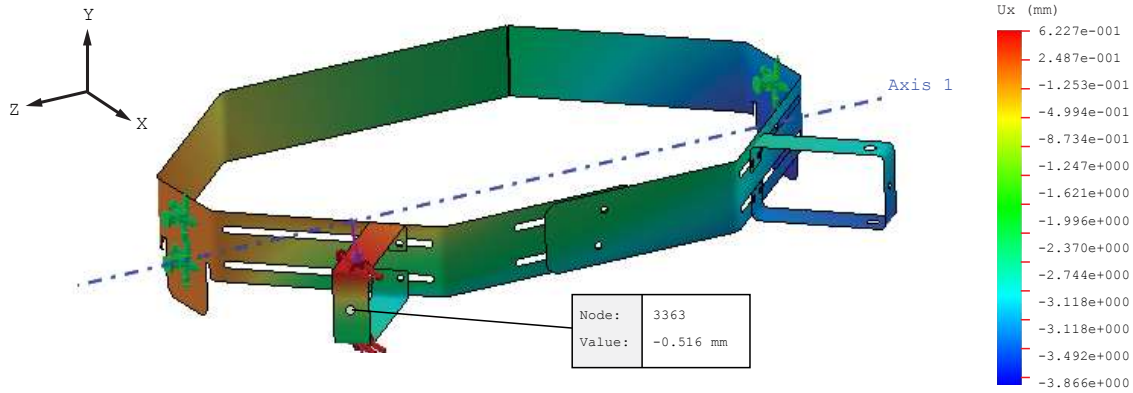


Figure 2.10 — Example of a FEA where the fixed pelvic structure is subjected to forces (red arrows) in the sagittal direction (z-direction) and moments (purple arrows) around Axis 1.

Next, deflections x_{PS} and x_{PF} in sagittal and frontal plane can be calculated and the new position of PA is calculated:

$$x_{sag} = \frac{F_x}{\kappa_{sag}} \quad (2.26)$$

$$x_{fro} = \frac{F_z}{\kappa_{fro}} \quad (2.27)$$

$$PA' = PA + [x_{sag}; 0; x_{fro}] \quad (2.28)$$

The same methods are applied to find the new coordinates of the TA.

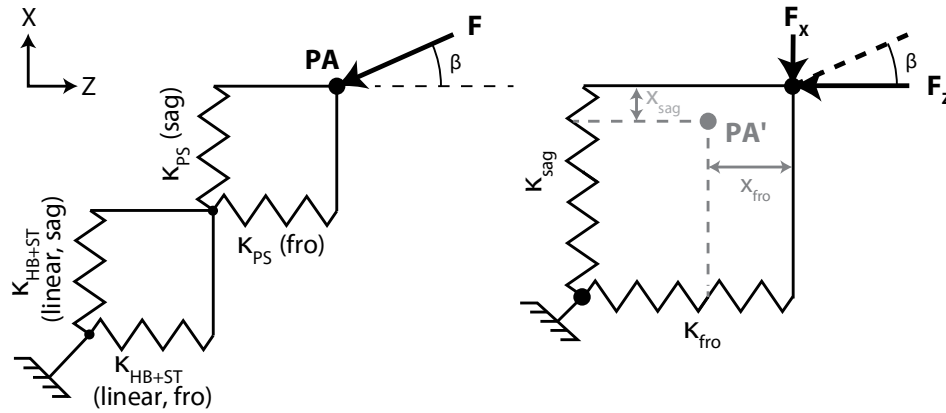


Figure 2.11 — New coordinates of PA in a 3D compliant environment. First equivalent stiffnesses are obtained, then new positions are calculated.

2.4 Optimization

The goal of this optimization is to minimize metabolic cost of walking by fitting a passive hip exoskeleton to the body. The orthoses will create advantageous moments around the HJC and therefore will alleviate the mechanical work of the subject's muscles around the hip joint.

Data used for the theoretical research is subtracted from an internal database from the University of Twente. Eight healthy able-bodied young adults with no known gait pathologies or pains volunteered for that study. Four male and four female students with an average age of 24.3 ± 1.4 years, height of 1.80 ± 0.1 m and weight of 68.8 ± 12.1 kg (mean value $\pm$ standard deviation) were recruited [16, 21].

2.4.1 Assumptions

1. While wearing the exoskeleton gait will not be altered by the wearer. With this assumption regular gait data can be used for optimization.
2. At the PA the leaf spring will remain vertically oriented, despite being subjected to substantial forces and moments at the PA.
3. Friction in the moving parts of the exoskeleton will be neglected.
4. For the optimization, metabolic cost due to added weight and inertia will not be calculated: the results for the optimization are an ideal representation.
5. The muscle model used for the optimization is rather straightforward with muscle efficiencies for both positive and negative work. Positive and negative mechanical work is purely delivered by muscles. A Hill-type model [46] of muscle representation with muscle force-velocity relations was not used and also the presence of tendons in the human body is not incorporated in this model.

2.4.2 Objective Function

The objective function J is to minimize the total metabolic energy in the hip joint of the human during gait and can be formulated from equation (2.16) as:

$$J = \min(E_{Meta}) \quad (2.29)$$

2.4.3 Constraints

During walking with the orthoses, the critical leaf spring may not encounter any behavior that cannot be described as elastic. Inherently, the spring may not plastically deform or fail when it is used. Therefore the maximum bending stress occurring at the outer layers of the clamped site cannot exceed the yield strength of the material σ_y . Together with a safety factor $sf = 1.1$, C_2 can be defined. Together with constraint C_1 (2.4) from the model this adds up to the following set of constraints:

$$\begin{aligned} C_1: & \quad d_4 < d_2 \\ C_2: & \quad \sigma_y \geq \frac{M_{Max} \cdot y}{I} \cdot sf \end{aligned} \quad (2.30)$$

2.4.4 Results

Excessive calculation time was needed to optimize the compliant model, hence the rigid model was optimized first, after which the best variables of this model are transferred to the compliant model. Successively, one variable that is known for increasing the stiffness of the system is chosen and iteratively increased until the compliant model reaches its optimum state. The variable chosen to increase the stiffness of the system is the radius of the circular cross-section of the leaf spring.

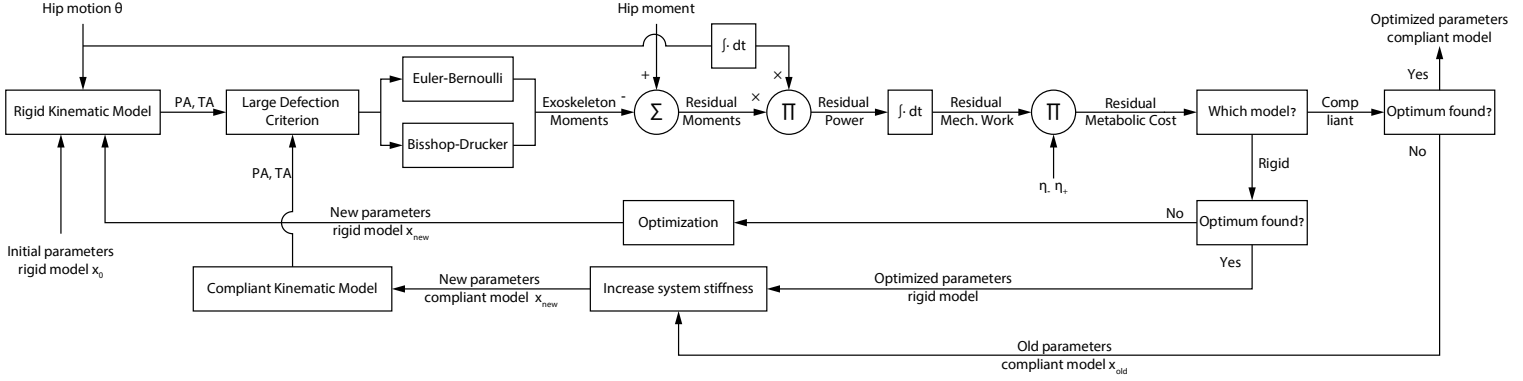


Figure 2.12 — Optimization process. Inputs are human motion and moments, outputs are the optimized parameters.

Furthermore, calculation time was reduced by using the Euler-Bernoulli beam theory for small angles and the more time-consuming Bisshopp-Drucker beam theory for larger angles. This is depicted in Figure 2.12 as the *Large Deflection Criterion*. Common finite element software also use criteria whether to solve for large displacements. SolidWorks (Concord, MA) uses the following criteria [47]:

$$\frac{\delta}{L_c} > 10\% \quad (2.31)$$

where the characteristic length L_c is defined as:

$$L_c = \sqrt{\sum_{i=1}^N ((X_i - X_c)^2 + (Y_i - Y_c)^2 + (Z_i - Z_c)^2) / N} \quad (2.32)$$

where X_i , Y_i and Z_i are the coordinates of node i , X_c , Y_c and Z_c are the coordinates of the geometric center of the model and N is the total number of nodes in the model. This can be interpreted as the average distance from the geometric center and can be represented as:

$$L_c = \frac{L}{4} \quad (2.33)$$

To find the most optimal values for the rigid model the cost function J (2.29) is optimized for with boundary conditions C_1 and C_2 as described in (2.30). Considering the possibility of different local minima in the solution space a MultiStart algorithm with fmincon as local solver

was used to find the global minima. Optimization was done with MatLab and a scheme of the process is shown in Figure 2.12.

Optimization results can be seen in Figure 2.13 and Table 2.2. A drop in peak torques can be seen in both planes, which in turn causes less power demand and a decrease in metabolic cost. Compared to normal gait, the metabolic energy expenditure for one full gait cycle drops to 26.0% (sagittal) and 82.6% (frontal) for the PHE compared to regular gait. In total the hip joint energy expenditure is 39.2% compared to regular gait, which can be seen in Table 2.3.

The reason for the frontal plane to decrease less in metabolic cost can be found in the quasi-static stiffness of the hip being around 5 times stiffer joint in the frontal plane compared to the sagittal plane.

Table 2.2 — Values for optimized parameters.

d_1 Distance [m]	d_2 Distance [m]	d_3 Distance [m]	r_1 Radius [m]	θ_1 Angle [rad]	d_4 Distance [m]	θ_2 Angle [rad]
0.1074	0.2000	0.3000	0.0037	1.4096	0.0936	1.4915

Table 2.3 — Resulting cumulative metabolic cost for one full cycle for sagittal, frontal and both (total) planes displayed as [J/kg] and percentages compared to regular walking.

	Sagittal [J/kg]	Compared to regular	Frontal [J/kg]	Compared to regular	Total [J/kg]	Compared to regular
Regular Gait	1.044	-	0.317	-	1.362	-
With PHE	0.272	26.0%	0.262	82.6%	0.534	39.2%

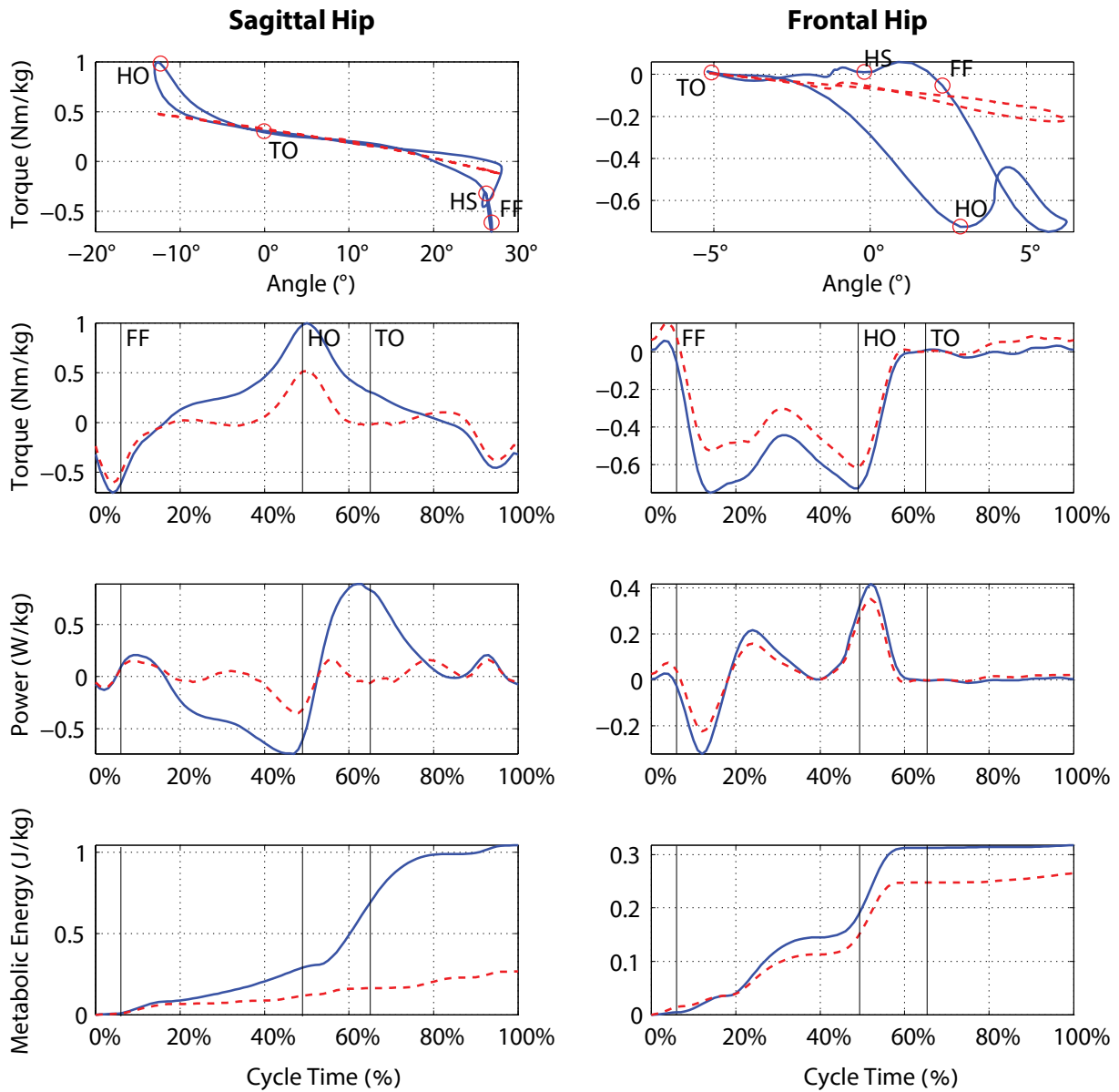


Figure 2.13 — Optimization results for one full gait cycle. Top row indicates the torque-angle characteristic of the human gait (blue) and the exoskeleton (dashed, red). In the second, third and fourth row the torque, power and metabolic cost are given for normal gait and their respective residuals for walking with the passive exoskeleton.

2.5 Design Concept

2.5.1 Material for Leaf Spring

The objective of the material can be described as to maximize the energy stored per unit weight, with constraints being that the material does not fail. The following material index M can thus be obtained:

$$M = \frac{\rho E}{\sigma_y^2} \quad (2.34)$$

where σ_y is the yield strength, ρ is the density of the material and E is the Young's Modulus of the material. Low values of material index M are favorable.

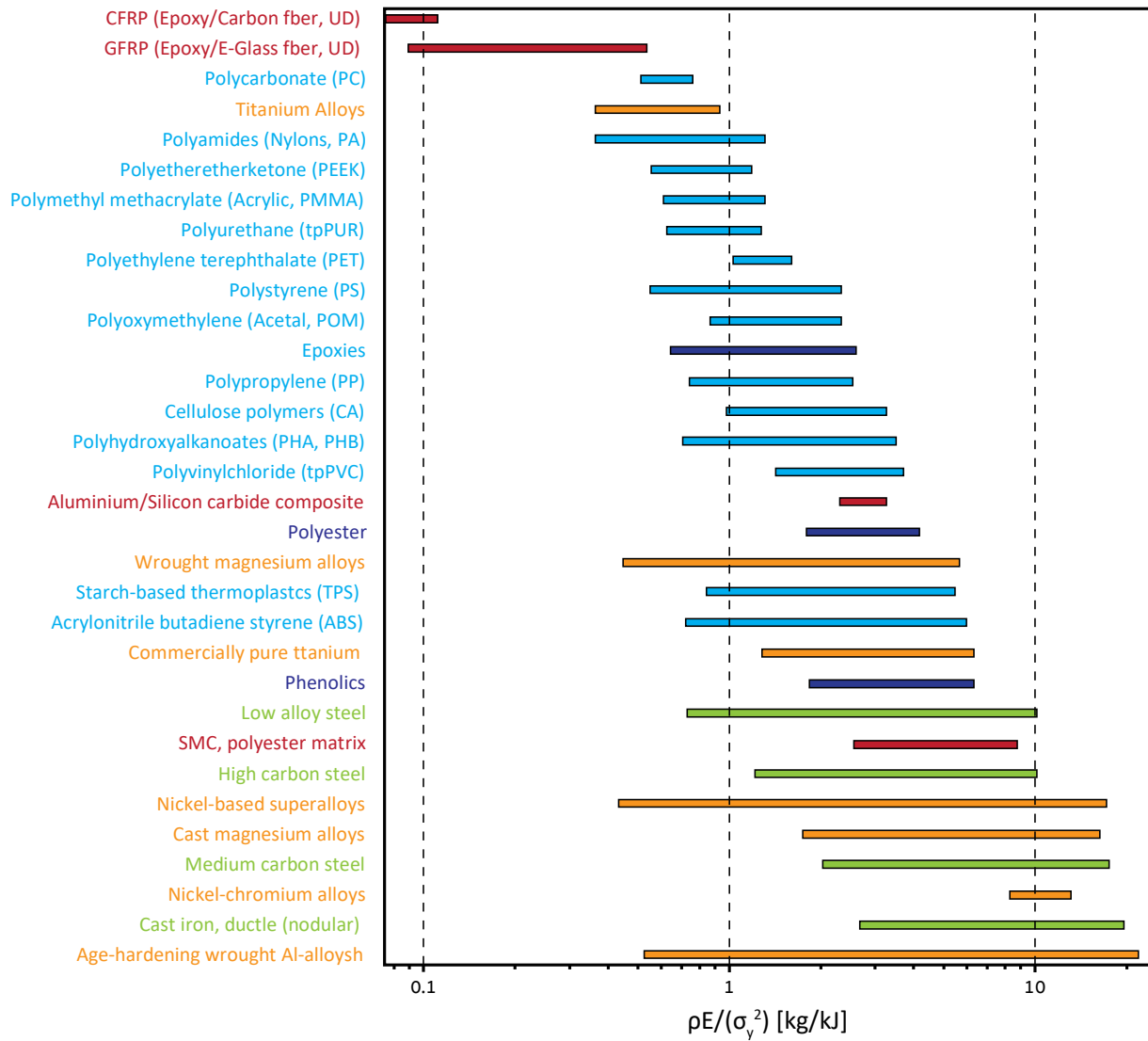


Figure 2.14 — A selection of materials with a minimum of 1 [GPa] for Young's Modulus plotted over material index M . Red: composites; light-blue: thermoplastics; orange: non-ferrous metals; dark-blue: thermosets; green: ferrous-metals.

A lower limit of 1 [GPa] is set for the materials to exclude very elastic materials, for which the leaf spring would be very thick to obtain at least some level of stiffness. An overview of materials with theoretical values for the material index M of equation (2.34) is given in Figure 2.14. Composites such as Carbon (CFRP) and Glass (GFRP) Fiber Reinforced Polymers score well, as well as a broad range of polymers including PC, PA, PEEK, PMMA, tpPUR, PET, PS and POM. The only metal which obtains high values for the material index M are titanium alloys.

In order to build a working prototype, unfortunately, availability in the Netherlands or delivery times to the Netherlands are limiting factors. Also, standard off-the-shelf products normally don't have the highest quality in terms of mechanical properties. Consequently, in order to choose a material, practical values, deducted from property sheets of manufacturers of the subsequent materials, have to be compared.

Table 2.4 — Materials of interest with low values of material index M . In bold the material names and in non-bold the distributors in the Netherlands.

	ρ [g/cm ³]	E [GPa]	σ_y [MPa]	$\frac{\rho E}{\sigma_y^2}$ $\left[\frac{\text{kg}}{\text{k}}\right]$
CFRP (Epoxy/Carbon fiber, UD, 0°)				
Van Dijk Pultrusion Products (DPP) [48]	1.59	140	1600	0.09
GFRP (Epoxy/E-Glass fiber, UD, 0°)				
Carbon Experience [49]	2.07	40	600	0.23
Polycarbonate (PC)				
Vink Kunststoffen B.V. [50]	1.20	2.5	70	0.61
Titanium alloys				
Titaniumshop.nl (Grade 5) [51]	4.42	120	1000	0.53
Polyamides (PA)				
Vink Kunststoffen B.V. [50]	1.14	3.0	80	0.53
Polyetheretherketone (PEEK)				
Vink Kunststoffen B.V. [50]	1.32	4.4	95	0.64
Polymethyl methacrylate (Acrylic, PMMA)				
Vink Kunststoffen B.V. [50]	1.19	3.0	74	0.65
Polyurethane (tpPUR)				
Vink Kunststoffen B.V. [50]	1.26	0.4	35	0.41

In Table 2.4 an overview of the top 8 materials from Figure 2.14 is given with values for ρ , E , σ_y and material index M as found in literature provided by manufacturers in the Netherlands.

Unlike the composites CFRP and GFRP, the polymers do have some differences compared to their theoretical values (for example #8 tpPUR has a lower value for M than #3 PC). A CFRP with an epoxy matrix and unidirectional fibers is favorable with the low material index M of 0.09.

Using fatigue strength to anticipate for many (typically 10^7 cycles) cyclic loads on the particular material would have been a good alternative in finding the most suitable material for this project. However, properties for fatigue strength are very rare to find on the internet. Most materials are subjected to a tensile bench test, to find values for yield, tensile strength etcetera. Fatigue tests are more time-consuming and costly and therefore values for these tests are rare to find for a variety of materials. Moreover, choosing the material based upon fatigue strength or yield strength normally will not differ much, since fatigue strength is related to yield strength. A safety factor will be introduced hereafter to anticipate on early failure.

2.5.2 Concept

The concept of the prototype is divided in the design of the thigh and the pelvic components.

Thigh Design

Since the thigh component is able to slide with respect to the leaf spring this means that the thigh component should be properly fixed to the body. Due to the downward conical shape of the upper leg, thigh cuffs that only act on the upper leg will slide downward due to gravity (Figure 2.15-A) and therefore are not sufficient.

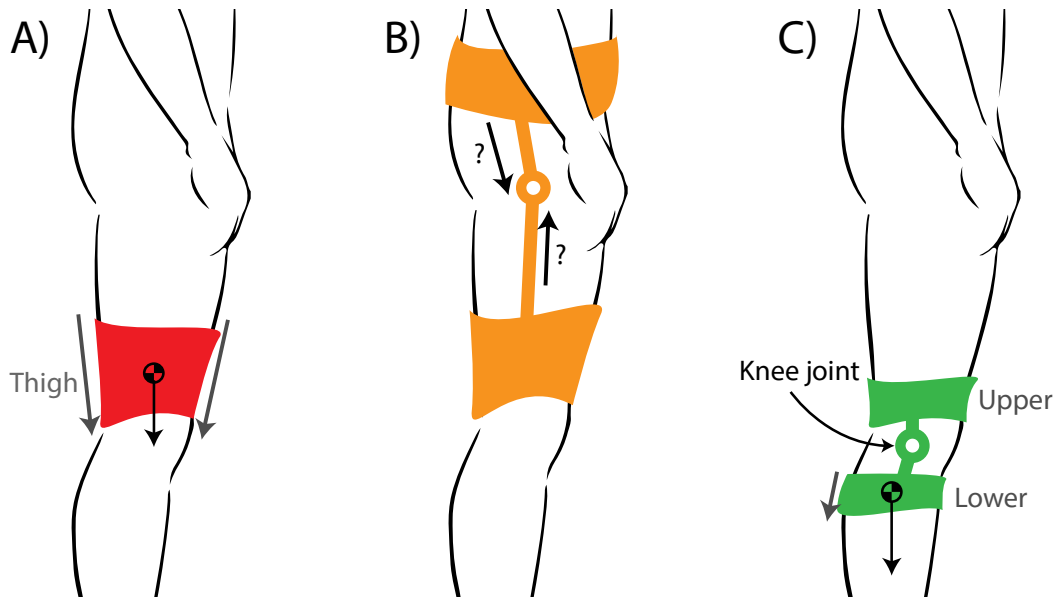


Figure 2.15 — Illustrations to show justification for the knee/thigh cuffs. **A)** Unfixed thigh cuffs will slide downward due to gravity. **B)** For a thigh cuff that is connected to the waist, aligning the mechanical axes with the human hip joint axes is difficult. **C)** A solution is an upper and lower knee cuff, for which the lower cuff is fixed relative to the body and the TA is attached at the upper cuff.

Another way to fix the thigh component is to rigidly attach the thigh brace with the pelvic brace (Figure 2.15-B). However it is difficult to properly align the mechanical joint axes with the human hip joint axes, moreover the thigh area has a large degree of soft tissue which is unfavorable (see subsection 2.3.5 Prototype and Human Compliance).

The area right beneath the knee has a smaller perimeter than the calf muscle so a cuff in this area will not slide down. However, since this part is rotating with the lower leg, an upper cuff has to be connected with the lower cuff through a mechanical revolute joint. In conclusion, the lower cuff is for fixing the thigh component to the body and the upper cuff is where the TA will be attached to.

For the attachment at the thigh, the following requirements apply:

1. TA should have 4 Degrees of Freedom
 - One translational DOF: along longitudinal axis
 - Three rotational DOFs
2. Should not create unwanted moments

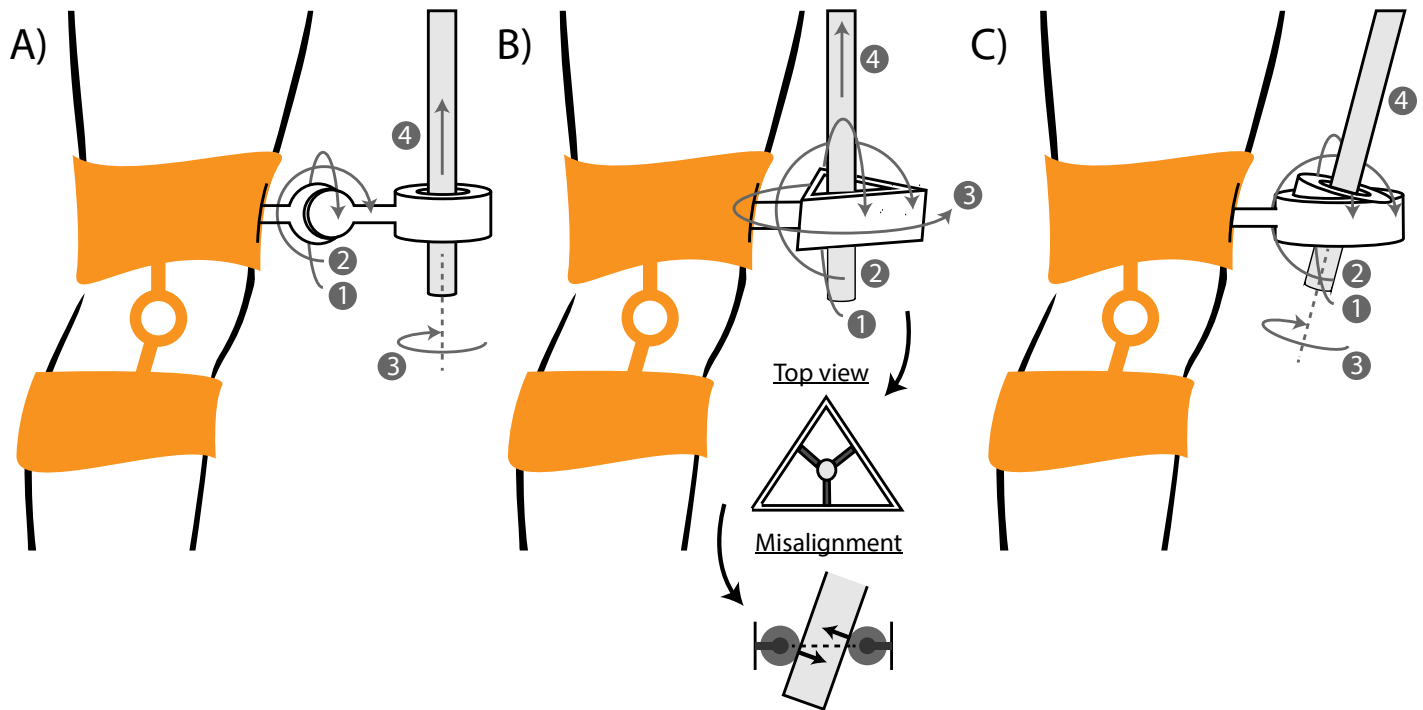


Figure 2.16 — Solutions for the connection between leaf spring and upper thigh cuff. **A)** Uses a ball-socket joint and a linear bearing, **B)** uses a mechanism with three perpendicular circular cams and **C)** consists of a spherical plain bearing.

In Figure 2.16 a small overview is given for particular ideas for the connection between the leaf spring and the upper thigh cuff: the TA. All of these concepts have four DOFs (depicted by the numbers 1 to 4). However, it is important that the axes of the DOFs in the TA are aligned, otherwise the rod will jam or set oblique which will create unwanted moments. In Figure 2.16-A

the ball and socket joint will inherently create the three rotational DOFs. Nonetheless the sliding linear bearing that will create the translational DOF is not aligned with the other DOFs. In Figure 2.16-B a mechanism is proposed which uses three circular cams whereas the leaf spring is the follower. On a macro scale, this mechanism will create the required four DOFs where all axes are aligned. However, on a micro scale (depicted at the bottom of Figure 2.16-B), misalignment occurs when the spring approaches with an angle. One way to align all DOFs and to evade unwanted moments is to make use of a combination of a spherical plain bearing together with a linear ball bearing, as depicted in Figure 2.16-C.

A model of the thigh component was 3D constructed (SolidWorks Simulation, Concord, MA) and can be seen in Figure 2.17. The two cuffs are connected by two revolute joints on both the lateral and medial side. On the upper cuff a construction is built to place the TA. This construction consists of a linear ball bearing press-fitted into a spherical plain bearing, which is subsequently attached to the plate construction. This construction also incorporates the possibilities for adjustments of the distance with respect to the HJC (variable d_3) and angle (variable θ_2). The cuffs are attached to the body by means of several Velcro straps/latches.

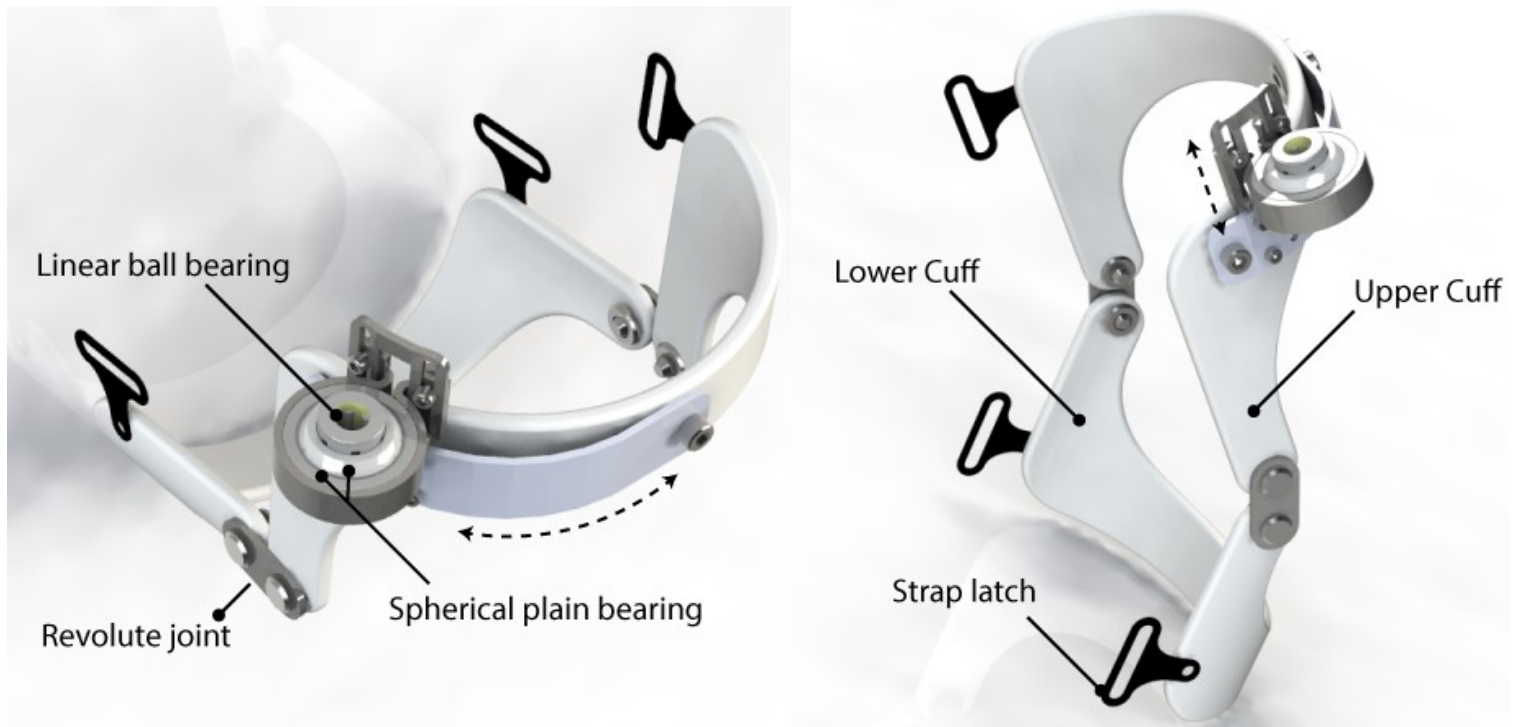


Figure 2.17 — Renders of the thigh component consisting of an upper and lower cuff. Connected with a revolute joint on both medial and lateral side. Four latches will strap the cuffs to the body. At the upper cuff a construction is built where the spherical bearing is housed and can be adjusted vertically and in the transverse plane.

The spherical bearing (INA, GE15-FW) allows up to 29° tilt according the manufacturer, while exceeding this tilt angle might introduce undesirable moments around the thigh. Therefore the arising tilt in the model, which is the difference between the femur angle (red dashed) and the

leaf spring angle (green dotted), is calculated during one gait cycle and can be seen as the purple dash-dotted line in Figure 2.18. It can be concluded that the tilt is not exceeding the physical properties of the spherical bearing (blue solid) at any point.

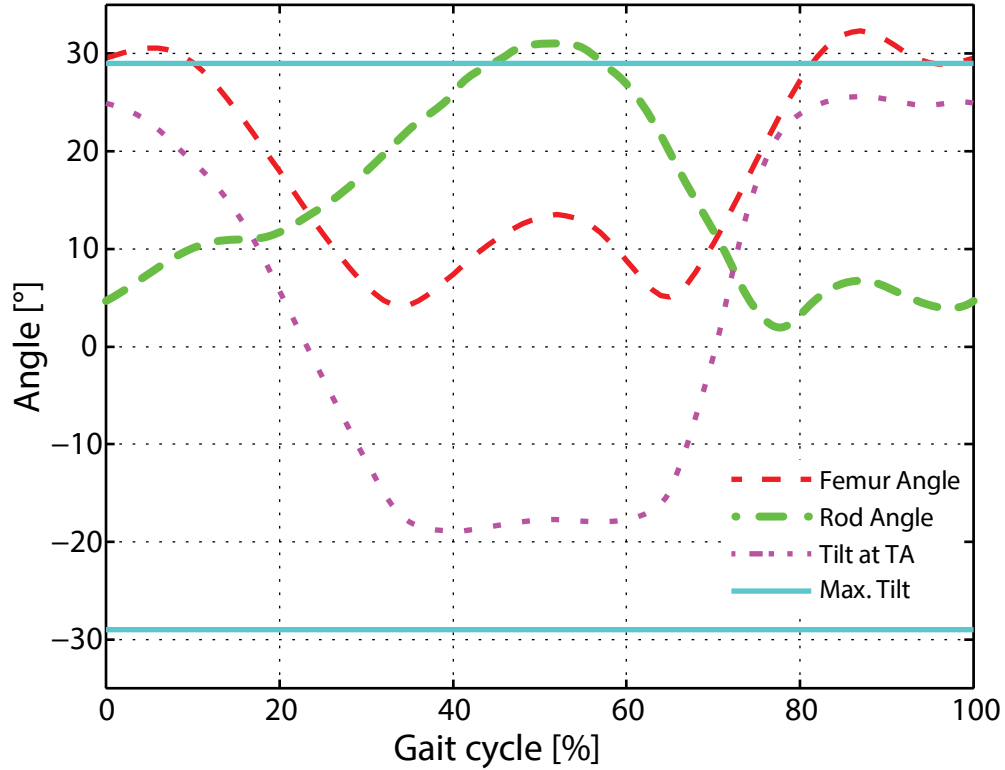


Figure 2.18 — Tilt at the TA (dash-dot, purple), calculated as the difference between the angle of the femur (dashed, red) and the angle of the leaf spring (dotted, green). The tilt angle at the TA never exceeds the physical properties of the selected spherical bearing (solid, blue).

Pelvic Design

For the pelvic design, a hip brace (Orthomerica, Newport 4 Hip Orthosis) is used, which has the advantage of two M6 nuts injection-molded into the brace. To achieve a higher robustness a steel structure was mounted around this brace, *Pelvic Structure* (Figure 2.19). The leaf spring is mounted onto this structure by means of a clamping mount.

The clamping site of the leaf spring can be altered by means of the following features (depicted as dashed lines in Figure 2.19):

- Slots in the *Pelvic Structure* (θ_1)
- Slots in the *Pelvic y-plate* (d_1)
- Position of the holes in the *Clamp Holder* (d_2)

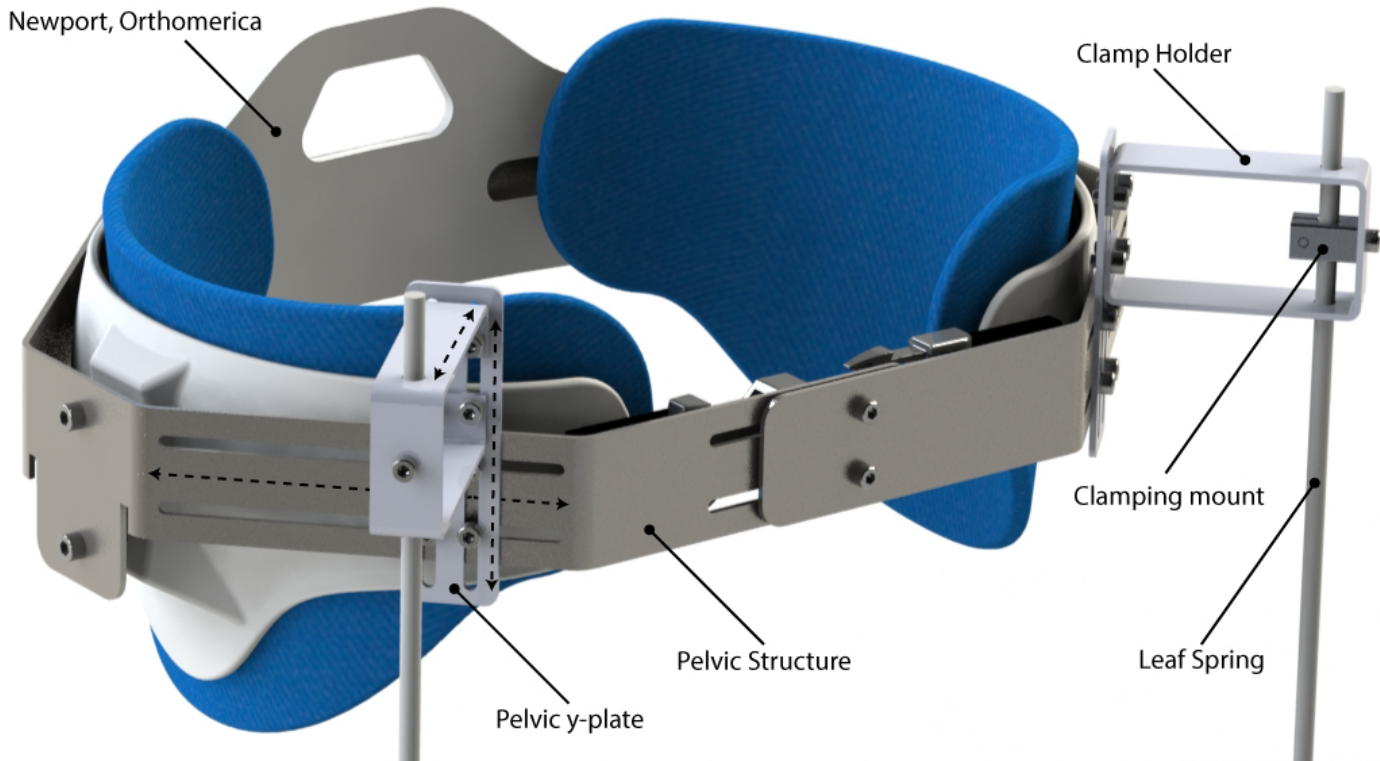


Figure 2.19 — Renders of the pelvic component. The hip brace (Orthomerica), a steel construction consisting of the ‘Pelvic Structure’, ‘Pelvic y-plate’ and the ‘Clamp Holder’. The leaf spring is fixed by a clamping mount onto the ‘Clamp Holder’.

2.6 Prototype

The prototype was manufactured at the students workshop at 3ME faculty at the Delft University of Technology.

2.6.1 First version

In Figure 2.20 an overview of the first prototype is given. The prototype is made adjustable in order to place the PA and TA according to the hip joint.

2.6.2 Second version

During the experimental trials with the first prototype, it was found that attachment between the Clamping mounts and the Pelvic y-plate was too compliant. Forces were not fully transmitted to the pelvic structure due to this compliance.

Another malfunction of the exoskeleton was caused by the cylinder, see Figure 2.20. A pair of very tiny set screws (M2x2) were inserted at the top and bottom of this cylinder to mount the cylinder onto the carbon leaf spring. However, the thickness of the cylinder was 1 mm and the set screw did not penetrate the rod fully. This caused the set screws to stick out of the cylinder,

blocking the linear ball bearing at full flexion or extension and resulted in a reduced range of motion.

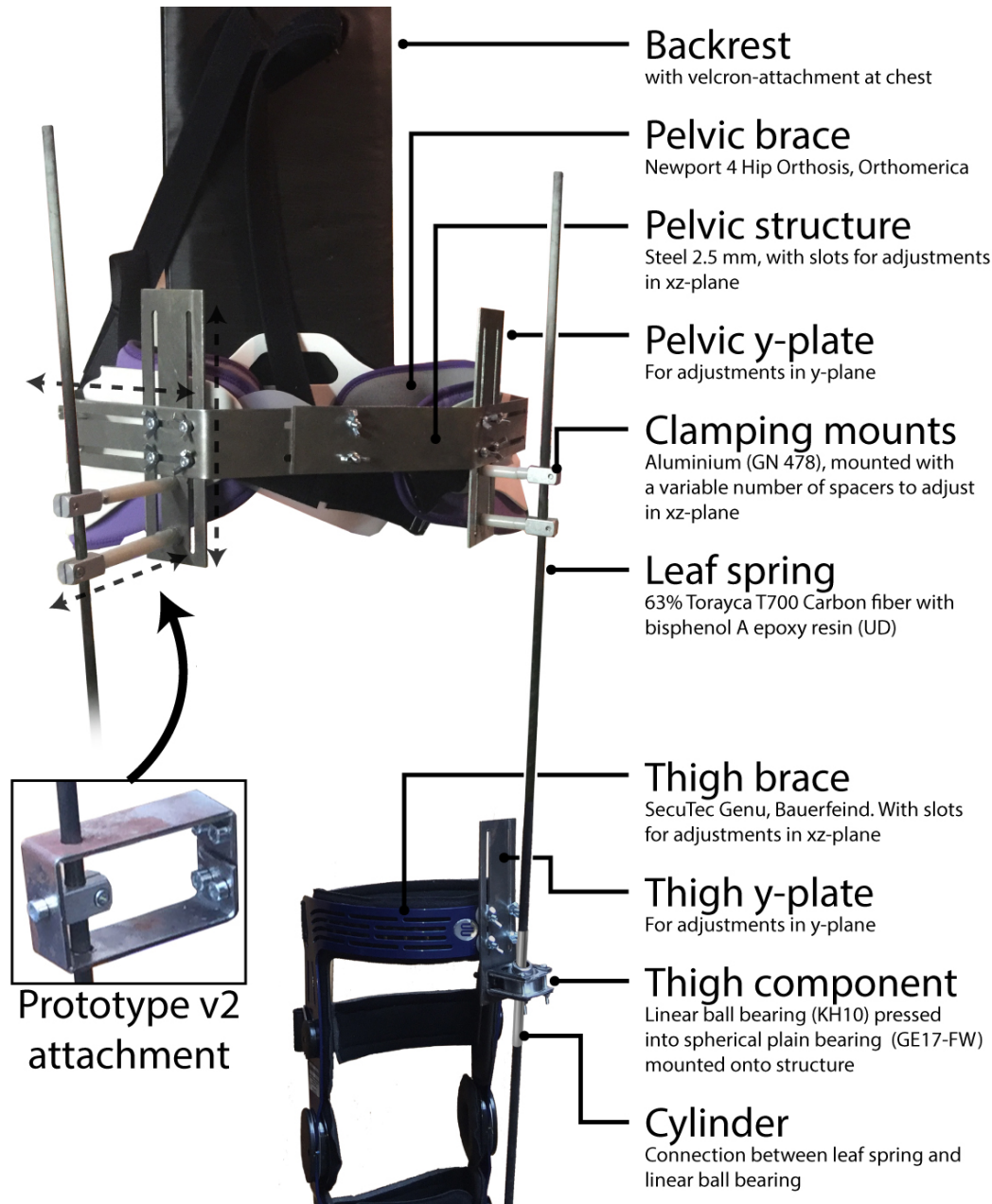


Figure 2.20 — Overview of prototype components. Inset: improvement of the PA, together with omitting cylinder results in second version of prototype.

In the second version of the prototype a new PA for the carbon leaf spring was manufactured, as seen in the inset of Figure 2.20. Also, it was decided to omit the cylinder. Increasing the range of motion was found more important than the introduction of some backlash at the TA.

The total weight of the prototype is 4.15 [kg] and can be subdivided into the waist part (2.55 [kg]), two thigh parts (each 0.78 [kg]) and two leaf springs (each 0.02 [kg]).

2.7 Experimental Evaluation

Treadmill gait trials were performed to evaluate the performance of the Passive Hip Exoskeleton. Goal of the experiments was to validate the hypothesis that for walking with the exoskeleton the energy cost of walking reduced. Electromyography (EMG) signals were measured around the left hip joint to gain more insight into which muscles are active during different phases of gait. To test the changes in metabolic cost of walking, oxygen ($\dot{V}_{O_2}$) and carbon dioxide ($\dot{V}_{CO_2}$) flow were measured. In order to do an inverse kinematics and dynamics analysis coordinates of the lower limbs were tracked and forces and moments on the treadmill were measured.

2.7.1 Subject

The author performed tests for this study, therefore no informed consent was required. Subject was healthy, 27 years old with weight 85 [kg] and length 1.83 [m]. Subject had no symptoms of gait disorder or any other physical disorder.

2.7.2 Apparatus and Measurements

1. *Muscle activity:*

Muscle activity data was recorded with a Delsys Bagnoli system (Delsys Inc., Boston, MA) using bipolar surface EMG electrodes and sampled at 1000 Hz. The following muscle groups were examined:

- a. Rectus femoris (hip flexion)
- b. Gluteus Maximus (hip extension)
- c. Adductor Longus (hip adduction)
- d. Gluteus Medius (hip abduction)

2. *Metabolic Cost:*

Measurements were conducted with a mobile Oxycon open-circuit spirometry system (CareFusion, Franklin Lakes, NJ). Average metabolic rates are reported every 5 seconds. A mask was worn to measure the gases and data was wirelessly transmitted to the host computer.

3. *Kinematics and Kinetics:*

To track the coordinates of the lower-limbs a Visualeyex system (PTI Phoenix Technologies, Vancouver, BC) with two sets of cameras running at 100 Hz was used. A split-belt force-sensing treadmill (M-Gait, Motekforce Link, Amsterdam, NL) running at 100 Hz was used to measure the ground reaction forces. Additional markers were placed on the exoskeleton to gain insight into the bending of the leaf spring.

A graphical representation of the experimental set-up and used apparatus for trials without exoskeleton 'NoExo', with exoskeleton but no springs attached 'ExoNS' and the complete exoskeleton 'Exo' can be seen in Figure 2.21.

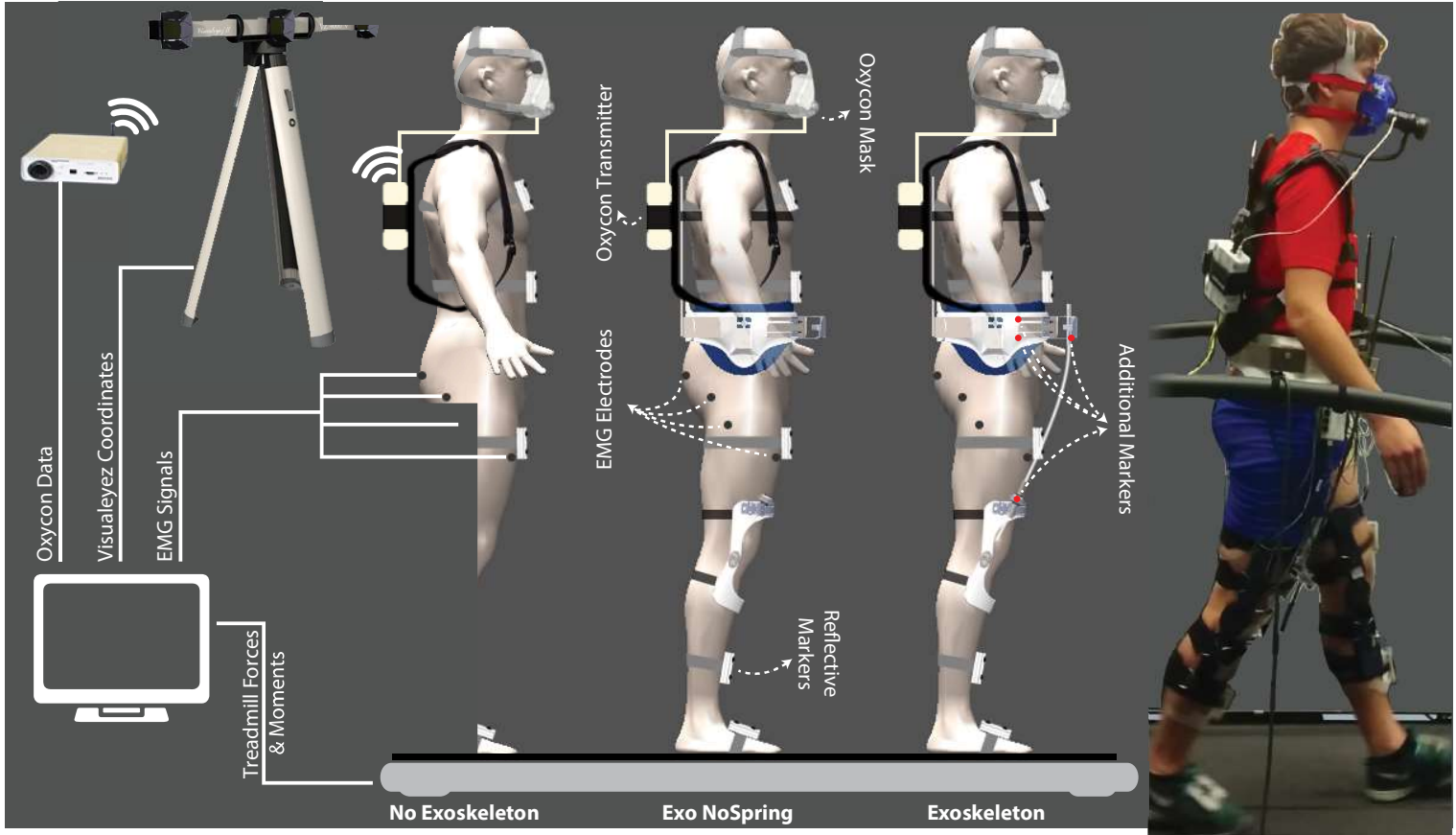


Figure 2.21 — Experimental setup displaying apparatus used for the different conditions. **Right:** screen capture from experiment.

2.7.3 Experimental Protocol

Preliminary to the walking trials, the subject's metabolic cost was measured while standing still. This 6 minute trial determined the metabolic steady state of the participant and the average of the last 5 minutes will be subtracted from the metabolic values during walking.

Next, the participant was prepared for the walking trials. The EMG electrodes for the Rectus Femoris, Gluteus Maximus and Gluteus Medius were placed according to the SENIAM guidelines [52]. The muscle belly location of the Adductor Longus was found midway between the origin at the pubic tubercle and the insertion at the medial linea aspera of the femur [53]. A reference electrode was attached to the kneecap, an area with no muscle activity. Reflective wireless marker groups, containing 3 reflective markers, were attached with straps to the feet, lower legs, upper legs, pelvis and trunk. A backpack containing a processing and transmitting device was worn by the user to gather gas flows originating from the face mask.

For trials containing the exoskeleton, the prototype was adapted such that the PA and TA were in the correct positions according to the HJC. To do this a non-invasive method was chosen to find the coordinates of the HJC. Kirkwood et al. [54] studied four non-invasive measures of

accurately finding the HJC and compared them to a standardized X-ray trial. The most accurate non-invasive method was by taking the midpoint of the line between the ASIS and the pubic symphysis (PS) and moving 20 mm inferior as described by Andriacchi and Strickland [55]. Referring to Figure 2.5, the coordinates of the HJC can now be calculated as:

$$HJC = \frac{ASI_x + PS_x}{2}; \frac{ASI_y + PS_y}{2} - 0.02; \frac{ASIS_z + PS_z}{2} \quad (2.35)$$

Thus by palpating and finding the coordinates of the ASIS and PS the coordinates of the HJC could be obtained using equation (2.35) and a 3D probe. Subsequently, PA and TA were positioned with respect to the HJC according to the optimized parameters using the same probe.

All trials were performed on a treadmill with fixed speed. Before the trials the subject chose its favorable speed, by incrementally increasing the speed of the treadmill. This self-selected speed was 1 m/s and matches the optimization condition. Another speed, 1.25 m/s, was chosen to study the effects of higher stride rate and muscular work.

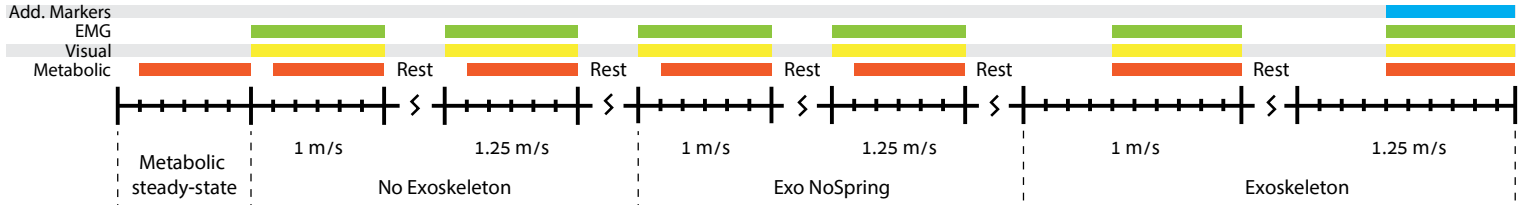


Figure 2.22 — Experimental protocol timeline. Subject completed a metabolic steady-state test to evaluate resting metabolic state before evaluating the three experimental conditions. In every condition (the last) 6 minutes of visual and EMG measurements was conducted. Metabolic values were obtained after (a minimum of) 1 minute walking. Every tick represents 1 minute.

2.7.4 Data processing

Processing of the analog data was done using Matlab (MathWorks Inc., Natick, MA, US). Data was cut up into sequences of individual stride cycles defined by the HS of the corresponding foot. HSs were detected using vertical forces of the treadmill. Kinematic and kinetic results and EMG data were normalized as a percentage of the stride cycle and averaged so that these graphs show one average stride cycle.

1. Muscle activity:

The analog EMG data was rectified and the DC offset was removed. A second order low-pass zero-phase Butterworth filter with a cutoff frequency of 50 Hz was used to remove noise and increase readability.

2. *Metabolic Cost:*

For energetics calculations, metabolic rates $\dot{E}$ (Watt/kg) were calculated with a formulae derived from [56]:

$$\dot{E} = \frac{16.48 \cdot \dot{V}_{O_2} + 4.48 \cdot \dot{V}_{CO}}{m} \quad (2.36)$$

Steady-state mean rates for $\dot{V}_{O_2}$ and $\dot{V}_{CO_2}$ were subtracted from mean values of every trial to yield net metabolic rate.

3. *Kinematics and Kinetics*

Marker data were filtered with a second order 20 Hz zero-phase Butterworth filter. The coordinates from the wireless marker groups to three bony landmarks of either limb were used for mapping out the kinematics of the subject. Landmarks used for the feet were the calcaneus and 1st and 5th metatarsal heads, for the lower leg the medial and lateral malleoli and the head of the fibula, for the upper leg the medial and lateral epicondyles of the femur and the greater trochanter, for the pelvis the anterior and posterior superior iliac spines and for the trunk the 7th cervical vertebra the xiphoid process and the Suprasternal notch.

Marker data and force data were synchronized by mapping a synchronization signal send by both systems.

Measurements of the treadmill resulted in forces and moments in three directions in the left and right belt. By using a scaled OpenSim [57] musculoskeletal model the external forces and moments of the left belt could be applied to the left foot and vice versa. This resulted in moments around the ankle which can be calculated up to the forces and moments around the trunk.

Chapter 3

Results

3.1 Theoretical

3.1.1 Beam Theories vs Finite Element Models

Reaction forces at the fixed PA of the Euler-Bernoulli and Bisshopp-Drucker model were compared with a finite element model shown on the right in Figure 3.1. A dynamic analysis was performed on the FE-model with the same CAD-software used before (SolidWorks Simulation, Concord, MA). A leaf spring with properties according to [48] was fixed at PA and had a no-penetration contact set at TA. A rigid structure had a hinge-like rotation at HJC and excited the leaf spring at TA. For this analysis only flex/extension rotations of the human data were used and compared with results given by the model presented in this study. Both (linear) Euler-Bernoulli and (nonlinear) Bisshopp-Drucker theories were tested and results can be seen in the left graph of Figure 3.1.

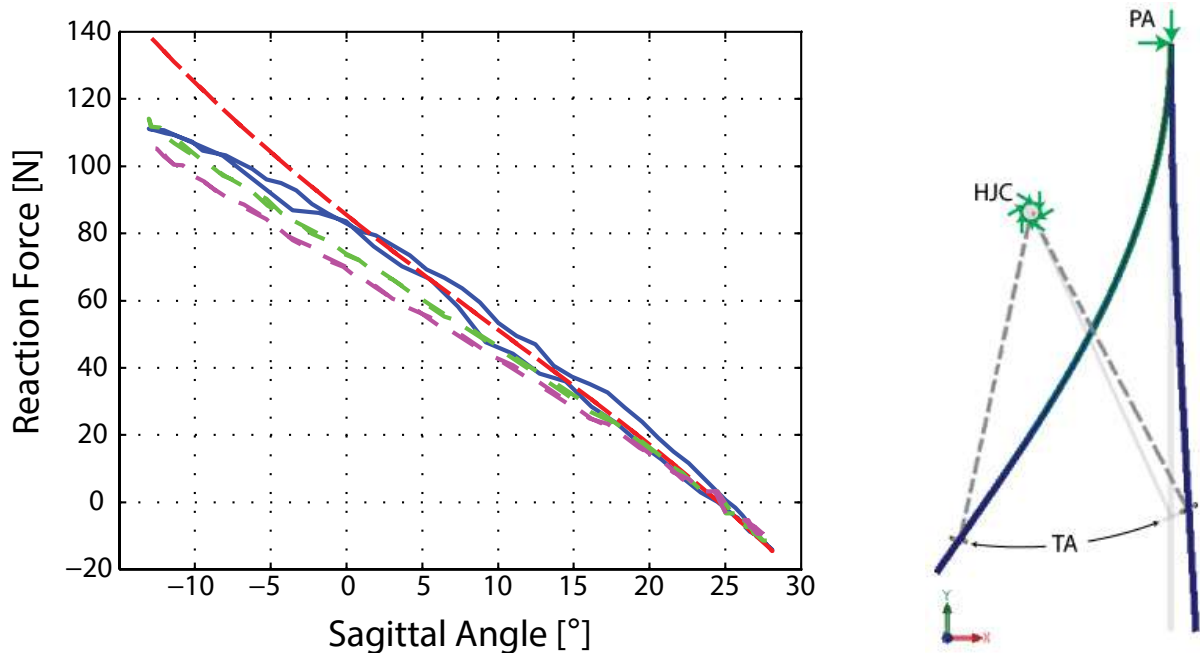


Figure 3.1 — Theoretical results comparing the different models. **Left:** Reaction forces at the PA for sagittal movements of the human hip during self-selected gait for the finite element model (blue solid), the (linear) Euler-Bernoulli theory (red dashed), the (nonlinear) Bisshopp-Drucker theory (green dashed) and the compliant model (purple dashed) presented in this study. **Right:** Corresponding FE-model with a fixed boundary condition at PA and a rotational fixture at the HJC, shown are results for maximum extension and flexion in the sagittal plane.

The linearity of the Euler-Bernoulli approach results in a higher reaction force at maximum extension compared to the nonlinear Bisshopp-Drucker theorem. The Bisshopp-Drucker theory is largely coincident with the FE-model. At maximum extension (-14°) and flexion (28°) the difference is approximately zero and maximum difference occurs at small angles with biggest difference to be approximately 10 [N].

3.1.2 Rigid vs Compliant Model

In the scenario of Figure 3.1 the results of the compliant model (purple dashed) are plotted as a reference to the rigid model (green dashed). Differences between the compliant and rigid model are smaller as expected, with 10 [N] in maximum extension.

3.2 Experimental

3.2.1 Muscle Activity

Compared to normal walking, EMG values increase when walking with the exoskeleton. One exception can be observed in the early stance of the rectus femoris muscle, where both conditions with the exoskeleton have a lower EMG value.

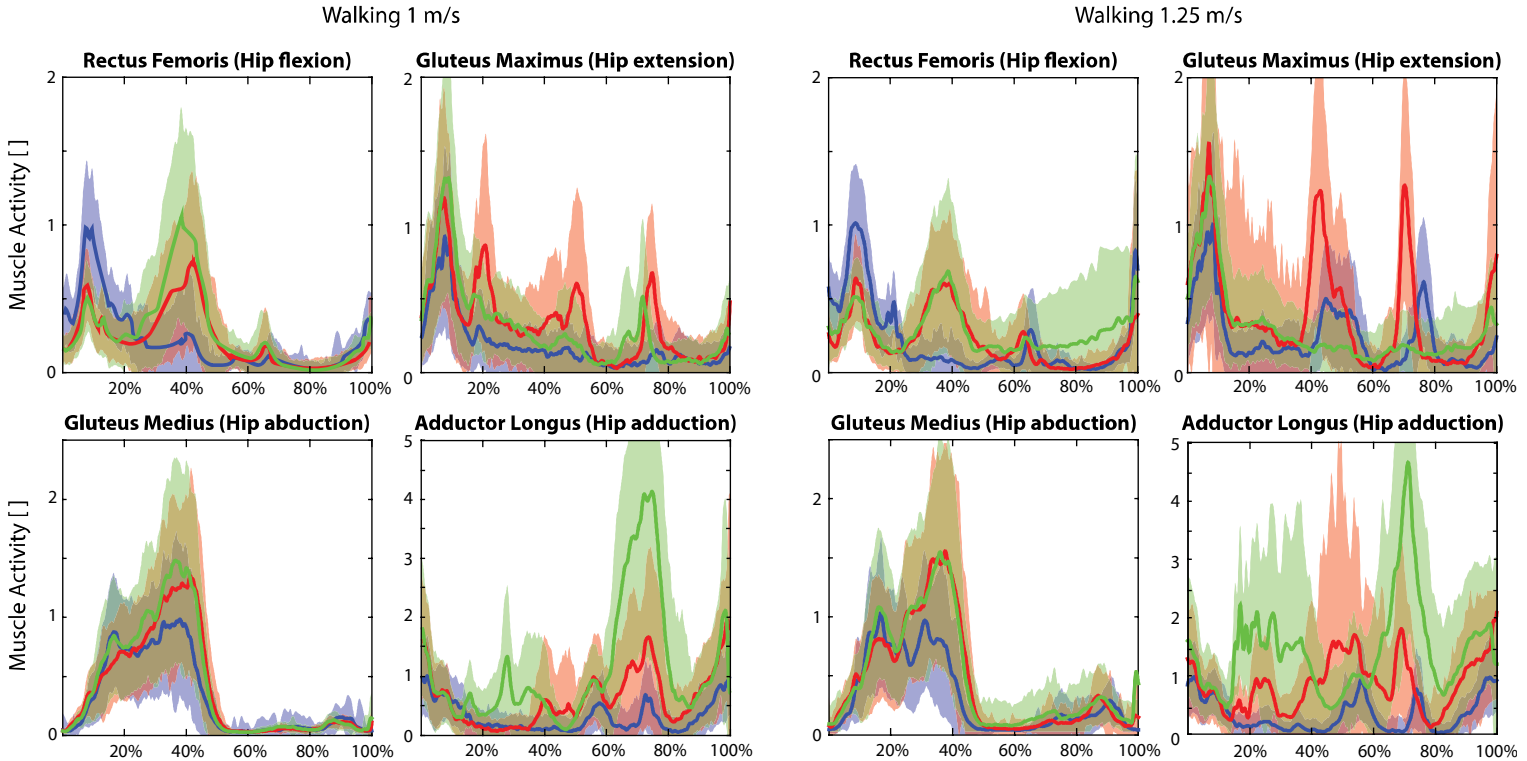


Figure 3.2 — Muscle activity measured with EMG electrodes for NoExo (blue), ExoNS (red) and Exo (green) at 1 m/s (**left**) and at 1.25 m/s (**right**). Signals are averaged for one stride starting and ending at HS and normalized from 0 to 1 for the normal walking condition. Solid lines are means and shades are plus and minus one standard deviation.

In the walking trials with the exoskeleton, conditions with and without the leaf springs are compared to determine the effect of the leaf springs. Decreases can be seen in the gluteus maximus throughout the whole stride in both walking speeds. In the rectus femoris increases can be seen during late stance and during swing but are only present in the slower and faster walking speeds respectively. Major increases are noticeable in the adductor longus for both walking speeds.

3.2.2 Metabolic Cost

Generally, when wearing the exoskeleton metabolic cost increases. Figure 3.3 shows the mean metabolic cost for the three different walking conditions. Results for the trials ‘NoExo’, 2.63 W/kg and 3.4 W/kg for walking at 1 m/s and 1.25 m/s respectively, are in accordance with other studies [10, 11, 15]. Compared to ‘NoExo’, walking with the inactive exoskeleton increases metabolic cost with 25.9% and 22.9% for walking at 1 m/s and 1.25 m/s respectively. When comparing ‘ExoNS’ with ‘Exo’ the metabolic cost increases with 11.8% and 0.5% for walking at 1 m/s and 1.25 m/s respectively.

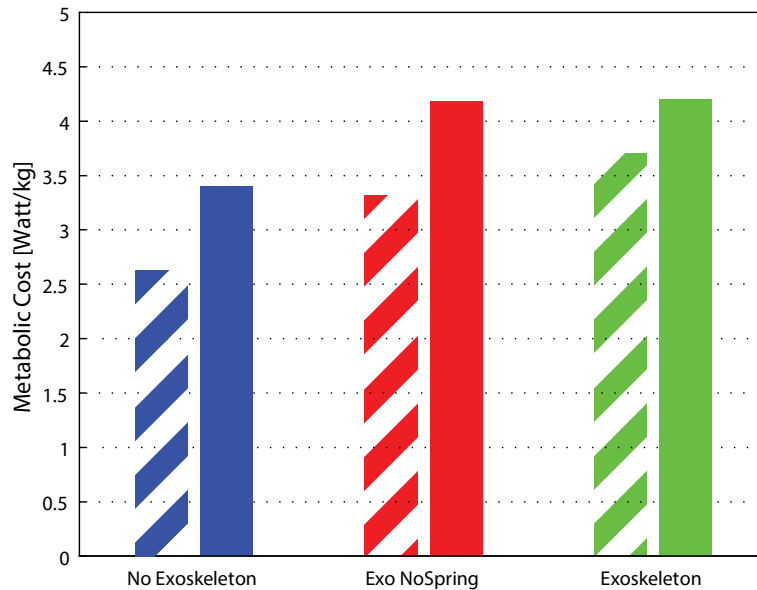


Figure 3.3 — Metabolic cost for the three different walking conditions for walking at 1 m/s (striped) and at 1.25 m/s (solid) for walking conditions ‘NoExo’ (blue), ‘ExoNS’ (red) and ‘Exo’ (green).

3.2.3 Kinematics

Please note that due to empty batteries for the wireless markers the results for ‘Exo’ at 1 m/s were unusable.

Wearing the inactive exoskeleton has a negative effect on the range of motion of the participant. At the knee this restriction of the range of motion can be seen around HS, FF, HO and during mid-swing. Around the sagittal hip this reduced range of motion is visible especially at late

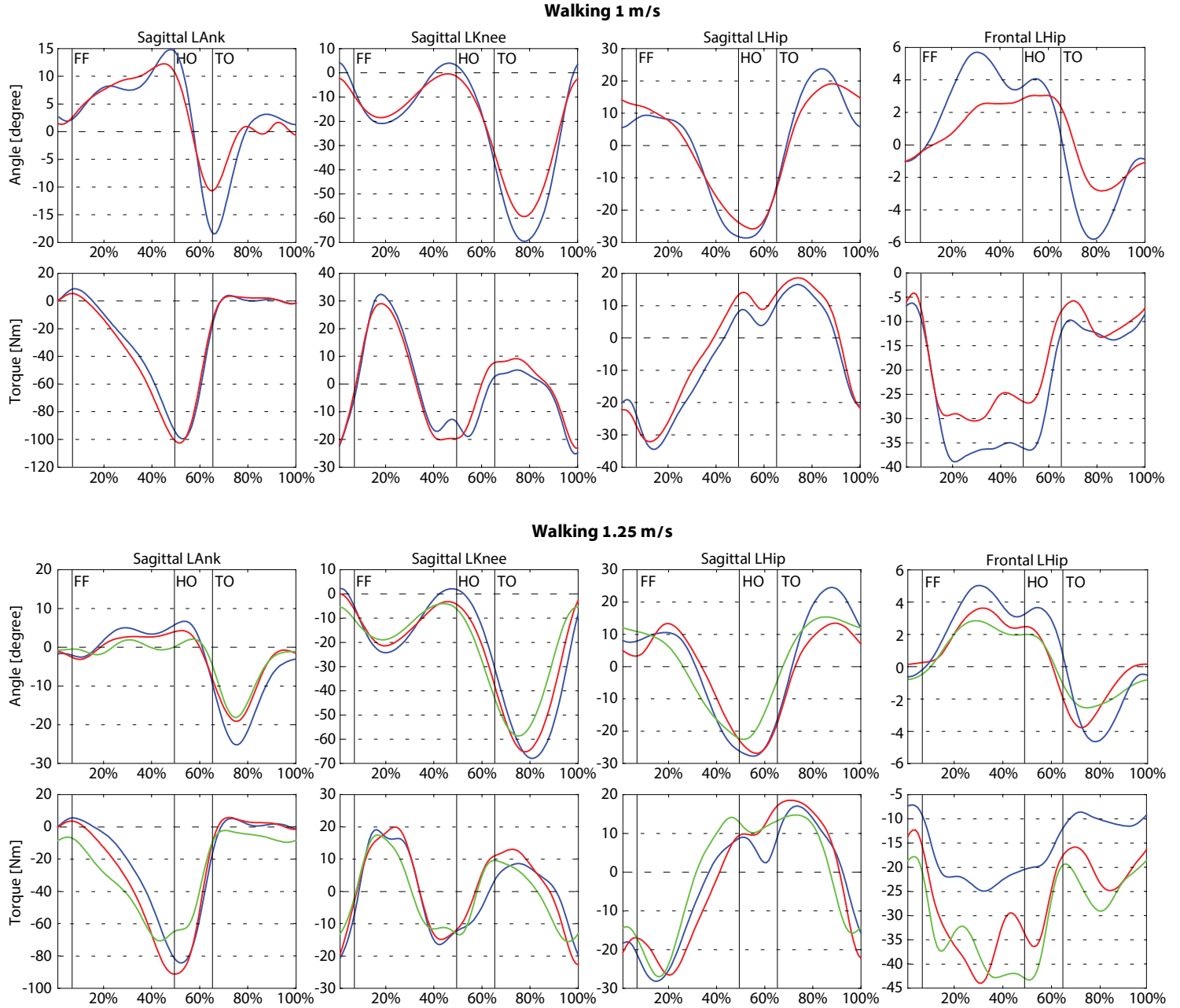
stance and late swing. At the ankle maximum plantarflexion is lower while wearing the exoskeleton. In the frontal hip plane, maximal abduction and adduction angles are reduced (Figure 3.4).

Comparing the exoskeleton with and without leaf springs, kinematics around the ankle are similar. At the knee, maximum flexion in swing is less when attaching the leaf springs. Maximum flexion of the sagittal hip is also reduced at late stance. In the frontal hip both abduction and adduction is decreased.

3.2.4 Kinetics

Generally, when walking with the inactive exoskeleton, the joint torques increase when compared to normal walking. Largest differences can be observed during powered plantarflexion of the ankle, maximum flexion in the knee and hip. In the frontal hip, however, joint torques decrease.

When comparing the inactive with the active exoskeleton, in general joint peak torques decrease. This can be seen around ankle plantarflexion, in all the maxima and minima of the knee and during begin and end of swing at the sagittal hip. In the frontal hip a general increase in joint torque is observed. Comparing the mean absolute torque of the sagittal hip we can see a decrease when wearing the active exoskeleton compared to the inactive skeleton. Furthermore a decrease in mean absolute torque is seen in the knee in contrary to the ankle and the frontal hip.



Mean absolute Torque [Nm]	Sagittal LAnk	Sagittal LKnee	Sagittal LHip	Frontal LHip
No Exoskeleton	22.21	10.01	12.59	16.00
Exoskeleton No Springs	26.61	10.82	14.63	27.09
Exoskeleton	27.40	9.20	13.08	30.65

Figure 3.4 — Kinematics and kinetics. Results shown in the top and bottom two rows are for walking at 1 m/s and 1.25 m/s respectively. Angles (top) and joint moments (bottom) are shown for ‘NoExo’ (blue), ‘ExoNS’ (red) and ‘Exo’ (green). Gait cycles start and end at HS. Table describes mean absolute torque at different joints for walking at 1.25 m/s.

3.2.5 Leaf Spring forces

Please note that additional marker data was only obtained during the 1.25 m/s walking trial.

During the exoskeleton trials, additional markers were placed onto the exoskeleton as is depicted in Figure 2.21. With the help of these visual marker sets insight was gained into the experimental bending of the leaf springs. In Figure 3.5 this experimental data is presented for the sagittal and frontal planes and is plotted against the theoretical outcome of the model.

In the sagittal plane, both prototypes give similar force shapes, nonetheless cannot produce the maximum force of 1.6 [N/kg] during late stance. However, it can be seen that the second prototype (dashed green) is stiffer and creates higher forces in both flexion and extension.

A rather unexpected force shape appears in the frontal plane. It can be seen that during stance limited abduction torques are produced to assist the human hip joint.

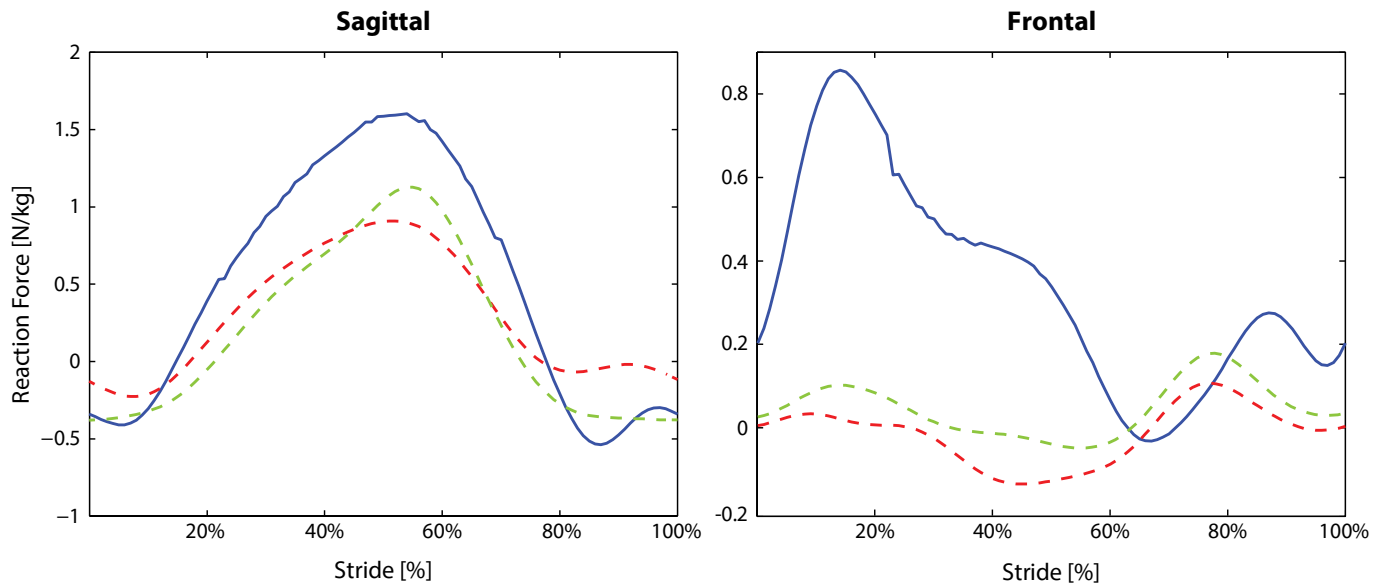


Figure 3.5 — Leaf spring forces deduced from visual information. Reaction forces at PA are shown for the predicted simulation (blue solid), experimental results for the first (red dashed) and second (green dashed) prototype version.

Chapter 4

Discussion

4.1 Muscle Activity

Walking with the inactive exoskeleton has a negative effect on muscle activity and generally increases the EMG values. An obvious reason for this can be the extra load worn on the body, but it can also be due to a stiffer walking pattern with more co-contraction caused by the uncertainty of the subject.

Big decreases are visible at the gluteus maximus during the complete stride. This is expected since the leaf springs alleviate muscle work. Furthermore at the second half of stance the hip is extending but flexion torques are exerted to achieve propulsion of the thigh into swing. Thus, when attaching the leaf springs, a decrease in muscle activity is expected at the second half of stance at the hip flexor rectus femoris, however is not seen. Yet, the rectus femoris is a biarticular muscle providing knee extension as well which might explain the increase. At the gluteus medius, decreases were expected, however instead increased slightly. Since hip adduction is not expected at all, surprisingly high EMG values arise at the adductor longus. This can be due to the additional role of the adductor longus as a hip flexor at the beginning of swing.

4.2 Metabolic Cost

Wearing the inactive exoskeleton increases metabolic cost, which can be easily explained by the addition of mass at the thigh and waist. Using the formulas of Browning et al. [19] we can calculate the influence of the added mass on the metabolic cost.

$$m_{thigh} = 0.78; \quad m_{waist} = 2.55;$$

$$0.075 \cdot (m_{thigh} \cdot 2) + 0.045 \cdot (m_{waist}) = 0.23 \left[\frac{W}{kg} \right] \quad (4.1)$$

This means only 0.23 [W/kg] is added to the metabolic cost of walking with the added mass of the exoskeleton. However at 1 m/s and 1.25 m/s the metabolic cost increases 0.68 [W/kg] and 0.78 [W/kg] respectively, comparing 'ExoNS' to 'NoExo'. Therefore approximately one third of the increased metabolic cost can be explained by the addition of weight. The other two thirds of the increased metabolic cost can be explained due to the reduced range of motion and the subsequent adapted gait. From kinematic and kinetic results a different walking pattern indeed is observed.

Wearing the active exoskeleton and walking at 1 m/s increases metabolic cost with 11.8% compared to 'ExoNS'. The increase is less than the 25.9% increase comparing 'ExoNS' with 'NoExo', nevertheless is still substantial. When comparing the active exoskeleton versus the inactive exoskeleton at 1.25 m/s, an increase is expected as well. However, no significant change is observed (+0.5%). This relative decrease might be the result of a learning effect, since the exoskeleton was first tested at 1 m/s after it was tested at 1.25 m/s (Figure 2.22). Another explanation of this relative decrease is explained in section 4.5 and relates to the system dynamics.

Furthermore, the ten minute breaks used in the protocol may be insufficient to avoid fatigue. Other studies use multiple-day breaks [20, 17] and a positive relation between distance covered and increase in metabolic cost of walking was shown in [58]. Spreading the experiments over multiple days giving the participant more rest, the metabolic cost of walking with the exoskeleton at 1.25 m/s might decrease compared to the inactive condition.

Despite these short-breaks and the use of motion-hindering braces this study already reached a better performance than other passive exoskeletons [13].

4.3 Gait Analysis

Wearing the exoskeleton has a negative effect on range of motion compared to normal walking. For example, in Figure 3.4 it can be seen that normally the subject has a small hyperextension at the knee joint, which can be seen during HS and HO. The knee brace (SecuTec Genu, Bauerfeind) used for the prototype contains an extension limitation at zero degrees, which means no hyperextension is possible. Furthermore flexion at the knee is reduced as well and can be caused by the tight straps or a misaligned mechanical joint. At the waist a hip brace (Newport 4, Orthomerica) is worn and again it can be seen that motion is hindered at the sagittal and frontal hip. Normally these braces are used to limit the range of motion of patients to prevent injuries or for rehabilitation, however in this study this property has an unwanted side effect. The same accounts for internal and external rotation at the knee and hip which are both prohibited by the respective braces and can lead to changes in gait.

The ankle joint is unhindered while wearing the exoskeleton, however range of motion still decreases when compared to normal walking. Hence, the limitations at the hip and knee might result into an adapted gait for which the result can be seen in reduced ankle angles while increasing ankle torques. Furthermore, to overcome the increased stiffness due to the motion-limiting braces the moments around the knee and hip are increased as well.

Attaching the leaf springs to the exoskeleton has an impact on the maximum flexion at the hip at late stance. This most likely has to do with perceiving resistance when tensioning the leaf springs. As a result the leaf springs are not maximally tensed, which also can be seen at the

sagittal reaction forces of the leaf springs (Figure 3.5). Training the participants to fully exploit the springs would result into a more efficient gait.

When wearing the active exoskeleton a decline in moments around the hip joint throughout the whole stride is expected, with maximum reductions around FF for extension and around HO for flexion (see the optimization results in Figure 2.13). Reductions around FF are indeed observed in the sagittal plane, however around HO an increase in flexion torques is seen followed by a decrease at TO. Throughout the whole stride mean absolute moments indeed have decreased wearing the active exoskeleton (13.08 [Nm]) compared to the inactive exoskeleton (14.63 [Nm]). Reductions in moments around the knee are observed especially around HS. A big reduction in plantarflexion moment is observed at the ankle, which might be an indication that the energy captured by the springs is used to lighten the work at the ankle joint instead of the hip joint to achieve forward propulsion of the body into swing.

4.4 Leaf Spring Forces

In the sagittal plane the maximum spring forces in extension are not met. Together with the reduction in the hip angles around maximum extension (Figure 3.4) it can be concluded that the user does not fully exploit the leaf springs.

Similarly a reduction in abduction and adduction angles can be seen in Figure 3.4. This might be influencing the frontal spring forces during stance, which compared to the predicting simulation are far from met.

4.5 Stride rate, natural frequency and pendular models

As described by Mochon and McMahon [59–60] the swinging limb at preferred walking speeds can be described as a pendular motion oscillating at its natural frequency. Metabolic cost is proven to be minimized around this natural frequency and is shown to increase with slower or faster frequencies [24]. The natural period during swing can be described as [59]:

$$T_{leg} = \pi \sqrt{\frac{I_l}{M_l \cdot g \cdot \bar{Z}}} = 0.49 \quad (4.2)$$

Using parameters of I_l , M_l and $\bar{Z}$ from Winter et al. [22] the natural period for the swinging limb for this study was 0.49 seconds. Assuming that the swing phase accounts for 38% of the total stride time and applying $f = 60/T$ the natural frequency during ‘NoExo’ becomes $47 [min.^{-1}]$. The exoskeletons knee brace’ centre of mass is equal to the centre of mass of the whole leg keeping inertial properties and therefore also the period unchanged when adding the exoskeleton. However, adding the leaf springs changes the dynamics of the total system since the springs have different dynamics:

$$T_{leg+spring} = \frac{\pi}{\sqrt{\frac{M_l \cdot g \cdot \bar{Z}}{I_l} + K}} = 0.30 \quad (4.3)$$

with K being the stiffness of the beam EI/L following Bernoulli's beam theory. This forces the subject to create a faster stride rate as shown by the green squares in Figure 4.1. A higher stride rate is achieved at 1.25 m/s compared to 1 m/s and therefore is closer to the natural frequency of the total system. Umberger et al. [24] showed that deviating from the preferred stride rate increases metabolic cost. This can explain the relative drop in metabolic cost for 'Exo' at 1.25 m/s as seen in Figure 3.3, since stride rate is closer to the natural frequency in this condition. However, since stride rate is limited by the velocity of the treadmill it is expected that metabolic cost will decrease further while walking at a higher velocity. When extrapolating a possible minimum can be found around 1.6 m/s (Figure 4.1).

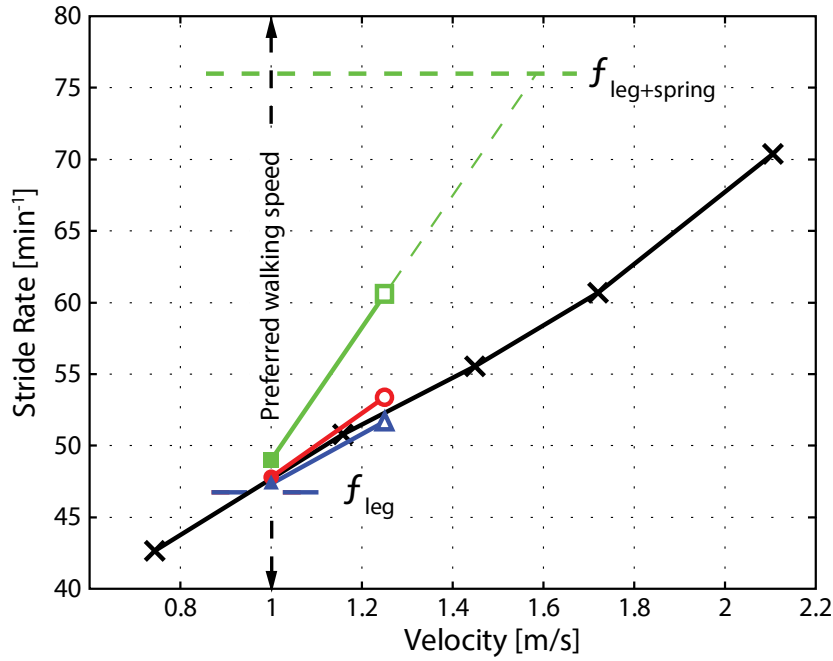


Figure 4.1 — Stride rate compared to velocity and corresponding natural frequencies. Stride rate versus velocity as defined in Minetti et al. [61] (black, **X**), during 'NoExo' (blue, **▲**), 'ExoNS' (red, **●**) and condition 'Exo' (green, **■**) for walking at the preferred walking speed (1 m/s, filled symbols) and at 1.25 m/s (open symbols). The natural frequency for 'NoExo' and 'ExoNS' based on the calculation in (4.2) is denoted with the lower horizontal dash (blue-red) and is applicable for self-selected gait. The upper horizontal dash (green) denotes the total system natural frequency.

4.6 Muscle-tendon interaction

In subsection 2.4.1 the assumption is made that mechanical work is purely delivered by muscles. However muscles are typically in series with a tendon, as denoted by the series element in Hill's model [46]. Several studies report a 50% contribution of the Achilles tendon to the positive

mechanical work delivered by the calf muscles due to elastic recoil of the tendon [62–63]. For this reason letting mechanical springs absorb energy during negative work phases of gait might interfere with the biological energy absorbing mechanisms. Hence it is often advised to only recycle energy that would otherwise be dissipated as heat, leaving phases of negative work followed by positive work unimpaired as these are likely for biological energy storage and release in the tendon [10]. On the other hand, muscles around the hip are relatively long and have short or no tendons and therefore the influence of passive tendon recoil will not be as severe as in the triceps surae-Achilles' complex, nonetheless might not be neglected.

4.7 Other

4.7.1 Comprehensive testing

In this study one participant for experimental validation was used. To gain more significant and reliable results more participants should be tested. Inter subject variability of gait parameters might be substantial: each individual has developed its own optimal strategy of walking.

4.7.2 Energy dissipation due to friction

Friction in the ball bearing might be a significant source of energy dissipation, however is shown to be negligible in the calculation below. Friction coefficient μ_k of the linear ball bearing according the manufacturer is 0.004, the average force $\overline{F_N}$ on the linear ball bearing is 40 [N], distance covered s during one gait cycle is 0.26 [m] and the average total stride time $\bar{t}$ is 1.165[s]. This gives the following result for the total energy loss rate due to friction in the linear ball bearing normalized to subject's weight W_b :

$$\dot{E}_{loss} = \frac{\mu_k \cdot \overline{F_N} \cdot s}{\bar{t} \cdot W_b} = \frac{0.004 \cdot 40 \cdot 0.26}{1.165 \cdot 85} = 0.42 \cdot 10^{-3} [W/kg] \quad (4.4)$$

This holds for approximately 0.01% of metabolic energy during 'Exo'. Sliding friction between the human and the exoskeleton is a more probable source of energy dissipation, however not tangible.

4.7.3 Transversal plane

Due to the leaf springs, internal and external rotations of the hip are to be expected, however are not included in the predicting model due to the relative small magnitude of these transversal moments around the hip. Assuming no sliding of the knee- and pelvic brace with respect to the femur and waist when imposed with these rotational forces, results for the transversal plane can be seen in Figure 4.2. It can be seen that predicted metabolic cost regarding transversal movements increases with 0.002 [J/kg] when wearing the exoskeleton. This is 0.3% compared to, for example, the reduction in metabolic cost of the sagittal hip in normal walking of

0.772 [J/kg] (Table 2.3) and hence is acceptably neglected. An unforeseen problem with these rotational moments however is that it might feel unnatural to the wearer. This can also be observed from the awkward torque-angle shape shown in Figure 4.2 and consequently can result in an adapted gait from the user refraining itself from internal and external rotations of the hip.

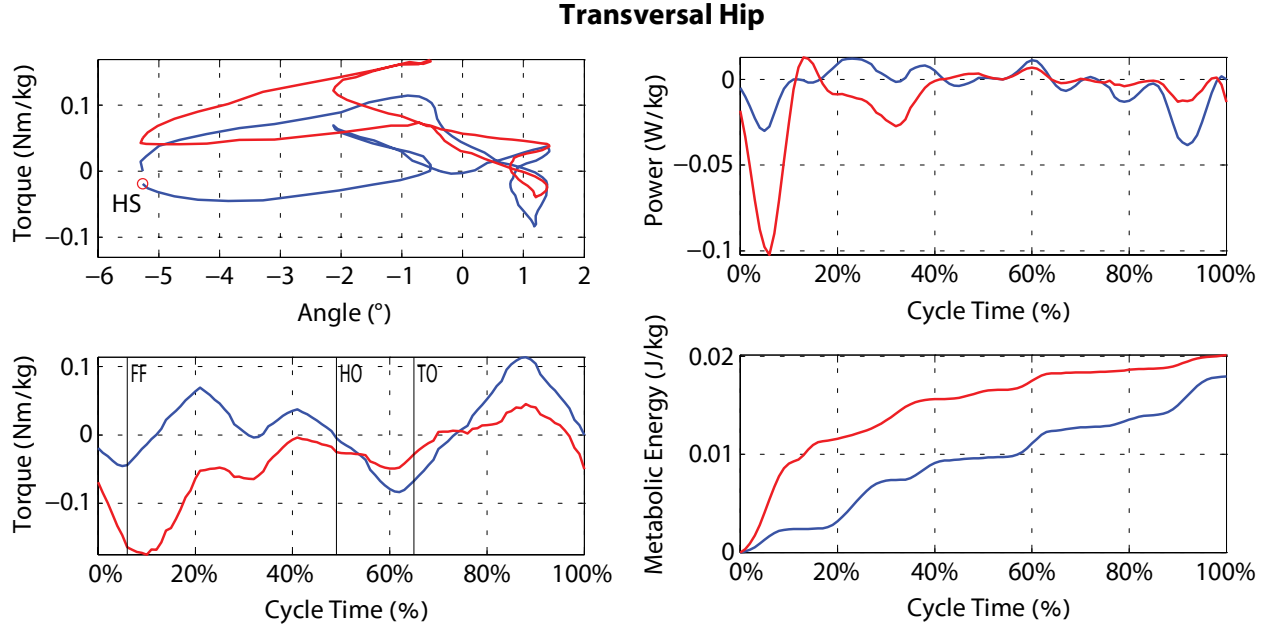


Figure 4.2 — A prediction of the effects of the exoskeleton on the transversal moments, power and metabolic energy.

4.7.4 Validating compliant model

Validating the compliant model with respect to the rigid model is difficult since the latter cannot be experimentally tested. However it would be useful to know to what extent the assumptions made in subsection 2.3.5 are in accordance with the compliance of the prototype and the soft tissue of the user.

One way to possibly gain insight into this matter is to look at the sagittal extension peak at late stance. As described in section 4.3, during this phase the user wearing the exoskeleton does not extend its leg as much as it would during normal walking. The reaction forces in the leaf spring would reach up to the orange dashed line in Figure 4.3, assuming the user would maintain its gait as in normal walking.

A fraction of the difference still persistent (Figure 4.3) might be due to an inaccuracy of the compliant model. Either magnitudes of the components of the compliant model are inaccurate or the model is incorrectly composed. Regarding the latter: in the compliant model it is assumed that the leaf spring is kept vertical at the PA as shown in Figure 4.4-A. However, it is very likely that the PA will get skewed by an angle $\Delta\theta$ due to the high forces imposed at TA. A skewed clamping site at PA has a negative effect on the deflection δ of the leaf spring, as shown

in Figure 4.4-B. Incorporating this in the compliant model would result into a bigger difference compared to the rigid model as shown in Figure 3.1.

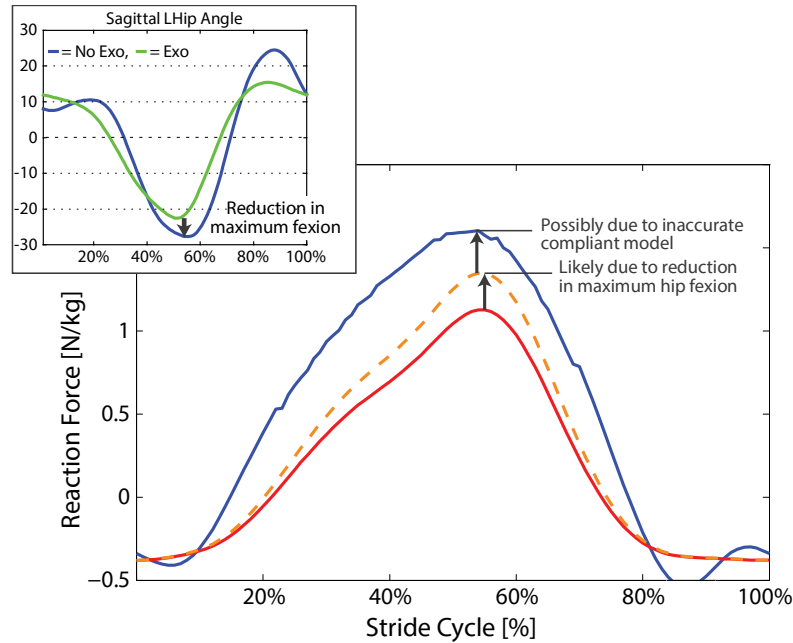


Figure 4.3 — Possibilities for mismatch optimization and experimental leaf spring forces. Reaction forces in the leaf springs as predicted from simulation (blue), as experimentally validated with the exoskeleton (2nd prototype, red) and as predicted when wearing the exoskeleton while maintaining gait as in normal walking (orange, dashed).

Analyzing the results of the motion capture system, this out-of-plane movement of the PA is indeed observed as can be seen in Figure 4.4-C and D. Maximum and minimum deflection being -0.016 and -0.037 and therefore a maximum difference in the position of PA along the y-axis of 0.021 [m]. This means that it is incorrectly assumed that ‘the clamped leaf spring at PA stays in a vertical upright position during gait’.

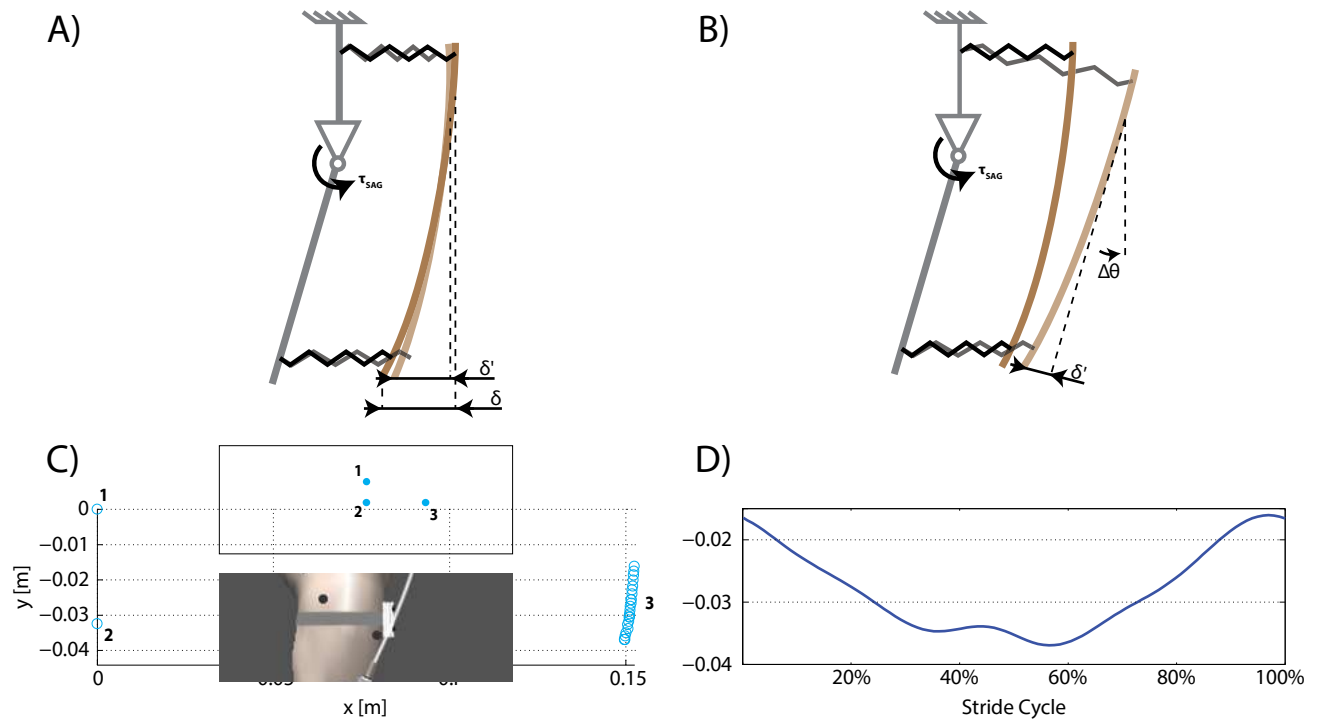


Figure 4.4 — Validating the compliant model . **A)** Model assumption: leaf spring remains vertical during gait. **B)** Prediction: leaf spring does not stay vertical and will be skewed by an angle $\Delta\theta$ affecting the magnitude of the deflection δ . **C)** Changes in the position of PA (3) with respect to the base of the structure (1 and 2) in the sagittal x-y plane. **D)** Changes in the position of PA during one stride cycle.

Chapter 5

Conclusion

In this study a Passive Hip Exoskeleton was designed, built and evaluated. The working principle is based on the fact that humans do negative work during gait, in which energy is lost in contrary to capturing and returning this energy with a mechanical spring. Theoretical optimizations show a maximum reduction of 60% of metabolic energy spent at the hip joint compared to regular gait by adding passive springs, without accounting for metabolic increases due to additional weight and changes in gait.

Gait trials with a prototype of the exoskeleton (total weight 4.15 [kg]) resulted in expected decreases in the moments exerted around the sagittal hip, confirmed by decreasing EMG muscle activity in the gluteus maximus. Metabolic cost with the exoskeleton compared to the inactive exoskeleton increased by 12% for walking at self-selected speed (1 m/s) and approximately remained equal for walking at 1.25 m/s. These values are a representation of total body metabolism and are difficult to compare with theoretic results for solitary hip metabolism. An explanation for the relative drop in metabolic cost for the higher walking speed is given by the change in natural frequency of the exoskeleton benefiting a higher stride rate and thus a higher walking velocity.

Furthermore, changes in the walking pattern were observed which can be attributed to the added distal mass and a decreased range of motion due to the exoskeleton. These changes in gait cause the user to not optimally make use of the leaf springs as simulated, decreasing the maximum forces exerted by the leaf springs.

The use of leaf springs parallel to the human leg has shown potential in passive exoskeleton design. Despite the usage of motion-constrained braces and short break intervals, the modest first prototype presented in this thesis already performed better than other passive exoskeletons. Therefore it is expected that with improvements to the prototype, suggested adaptations to the model and a more comprehensive experimental protocol, significant achievements for the presented approach will be obtained concerning the reduction of energetic cost in human walking.

Chapter 6

Recommendations

Changes to the exoskeleton

A few changes to the exoskeleton might have significant impact on the results of this study. Increasing the range of motion is a key point and the current study showed that the usage of rehabilitation braces is undesirable. A better approach would be to have custom-made attachments to the pelvis and knee allowing for flex/extension, ab/adduction and internal/external rotation without losing robustness.

To shift the focus to the original hypothesis to reduce metabolic cost compared to normal walking, improvements have to be made regarding the weight of the exoskeleton. The current weight of 4.15 [kg] is too heavy and especially the knee cuffs with a total of 1.56 [kg] are a substantial burden on energy expenditure. Therefore lightweight materials should be used for the pelvic and thigh braces without losing stiffness to bear the moments around the hip.

Influence on gait

Future research should also focus on how this particular exoskeleton (or exoskeletons in general) influences the gait kinematics and kinetics. In the current study it is assumed that gait kinematics do not change, nevertheless it is shown that this is not the case.

Gait trials with higher treadmill velocities

It is recommended to perform gait trials at higher treadmill velocities to study the influence of stride rate on metabolic cost. In section 4.5 it is predicted that walking at higher velocities with the exoskeleton should allow the user to reach the natural frequency of the exoskeleton system and would therefore minimize metabolic cost.

If results indeed show that reaching the natural frequency of the total system minimizes metabolic cost it should be investigated how the passive system can be adapted to fit different gait speeds. Increasing the effective length of the parallel springs will lower the natural frequency of the total system, however will reduce the aiding moments around the hip. Presumably, a trade-off between the two has to be made.

Chapter 7

Appendices

7.1 Appendix 1: Bisshopp-Drucker Beam Theory

A modification of the classical beam theory was presented by Bisshopp and Drucker in 1945 [33]. This new theory can be used for large deflections of cantilever beams and takes into account the shortening of the moment arm, see Figure 2.6-bottom. It is assumed that the bending does not change the length of the beam (inextensible) and the Euler-Bernoulli assumption that states that the curvature is proportional to the bending moment is still applied. Bisshopp-Drucker state that the curvature of the elastic line can be expressed in terms of arc length s and slope angle ϕ .

$$\frac{d\phi}{ds} = \frac{P}{EI}(L - x - \Delta) \quad (7.1)$$

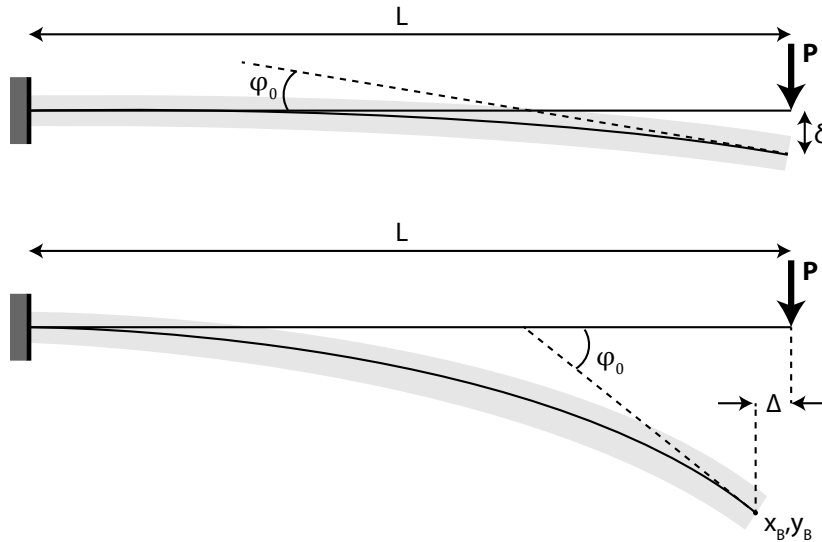


Figure 7.1 — Euler versus Bisshopp, linear versus nonlinear. Euler-Bernoulli beam theory (**top**) and Bisshopp-Drucker nonlinear beam theory (**bottom**). The clamped beam has length L with a vertical force P at the beam end, causing a deflection δ and slope of the bended beam ϕ_0 . The Bisshopp-Drucker theory incorporates an extra term Δ : the horizontal component of the displacement of the loaded end of the beam.

Manipulating equation (7.1) gives:

$$EI \frac{d\phi}{ds} = P(L - x - \Delta) = M \quad (7.2)$$

Calculating the derivative of equation (7.1) gives:

$$\frac{d^2\phi}{ds^2} = -\frac{P}{EI} \frac{dx}{ds} = -\frac{P}{EI} \cos \phi \quad (7.3)$$

Integration gives:

$$\frac{1}{2} \left(\frac{d\phi}{ds} \right)^2 = -\frac{P}{EI} \sin \phi + C \quad (7.4)$$

With boundary conditions being zero beam rotation at the clamped site $\phi(0) = 0$ and no curvature at the loaded end $\frac{d\phi}{ds}(L) = 0$ constant C can be calculated:

$$C = \frac{P}{EI} \sin \phi_0 \quad (7.5)$$

Inserting constant C into equation (7.4):

$$\frac{1}{2} \left(\frac{d\phi}{ds} \right)^2 = -\frac{P}{EI} \sin \phi + \frac{P}{EI} \sin \phi_0 \quad (7.6)$$

Manipulating equation (7.6) gives the curvature $\frac{d\phi}{ds}$:

$$\frac{d\phi}{ds} = \sqrt{\frac{2P}{EI}} (\sin \phi_0 - \sin \phi)^{\frac{1}{2}} \quad (7.7)$$

Now, its inverse becomes:

$$\sqrt{\frac{2P}{EI}} ds = \frac{d\phi}{\sqrt{\sin \phi_0 - \sin \phi}} \quad (7.8)$$

Due to the beams inextensibility, ds in the above equation can be integrated from 0 to L :

$$\sqrt{\frac{2P}{EI}} \cdot L = \int_0^{\phi_0} \frac{d\phi}{\sqrt{\sin \phi_0 - \sin \phi}} \quad (7.9)$$

To evaluate this elliptic integral, α is denoted by $\sqrt{PL^2/EI}$ and:

$$1 + \sin \phi = 2k^2 \sin^2 \theta = (1 + \sin \phi_0) \sin^2 \theta \quad (7.10)$$

Implementing (7.10), equation (7.9) can be written as:

$$\alpha = \int_{\theta_1}^{\frac{\pi}{2}} (1 - k^2 \sin^2 \theta)^{-\frac{1}{2}} d\theta \quad (7.11)$$

with $\sin \theta_1 = \sqrt{2}/2k$.

The elliptic integral at the right side of equation (7.11) can be written as the difference of two integrals of the first kind, complete and incomplete, using the notation of Jahnke and Emde [64]:

$$\alpha = F(k) - F(k, \theta_1) = \int_0^{\frac{\pi}{2}} \frac{d\theta}{\sqrt{(1 - k^2 \sin^2 \theta)}} - \int_0^{\theta} \frac{d\theta}{\sqrt{(1 - k^2 \sin^2 \theta)}} \quad (7.12)$$

where, from (7.10):

$$k = \sqrt{\frac{1 + \sin \phi_0}{2}}$$

$$\theta = \arcsin \frac{1}{\sqrt{2}k}$$

Values of $\sin \phi_0$ range from 0 (unloaded beam) to a maximum angle of 90° and so parameter k should vary between:

$$\sqrt{\frac{1 + \sin 0}{2}} < k \leq \sqrt{\frac{1 + \sin 90^\circ}{2}} \quad (7.13)$$

Now X_B can be described in two ways:

- 1) Via Bisshopp [33], eqn. (10) p. 275:

$$X_B = L \sqrt{\frac{2 \sin \phi_0}{\alpha}}$$

- 2) Via the kinematic model described in subsection 2.3.2:

$$X_B = |TA_y - PA_y|$$

Equalizing these two X_B 's gives the force P as a function of k . Placing this function for P back into the given $\alpha = \sqrt{P(k)L^2/EI}$ and using equation (7.12) the error between these two α 's can be calculated by iteratively increasing parameter k from $\sqrt{1/2}$ to 1. When a desired tolerance is reached, a value for k has been found and subsequently can be used to calculate the force P .

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