Handle With Care

The Anterior Hip Capsule Plays a Key Role in Daily Hip Performance

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Background: Passive energy storage and return has long been recognized as one of the central mechanisms for minimizing the energy cost needed for terrestrial locomotion. Although the iliofemoral ligament (IFL) is the strongest ligament in the body, its potential role in energy-efficient walking remains unexplored.

Purpose: To identify the contribution of the IFL to the amount of work performed by the hip muscles for normal, straight-level walking.

Study Design: Controlled laboratory study.

Methods: Straight-level walking of 50 healthy and injury-free adults was simulated using the AnyBody Modeling System. For each participant, the bone morphology and soft tissue properties were nonuniformly scaled. The superior and inferior parts of the IFL were represented by 2 springs each, and a linear force-strain relation was defined. A parameter study was conducted to account for the uncertainty surrounding the mechanical properties of the IFL. The work required from the gluteus, quadriceps, iliopsoas, and sartorius with and without inclusion of the IFL was calculated. Analysis of variance with subsequent post hoc paired *t* test was used to test the significance of IFL presence on the required mechanical work.

Results: During walking, the strain in the IFL reached a median of 18.7% (95% CI, 8.0%-26.5%), with the largest values obtained at toe-off. With the IFL undamaged and fully operational, the effort required by the hip flexor muscles was reduced by a median of 54% (99% CI, 45%-62%) for the iliopsoas and by a median of 41% (99% CI, 27%-54%) for the sartorius muscles. The inclusion of the IFL did not significantly alter the work required by the gluteus and the quadriceps.

Conclusion: The findings emphasized the key role the IFL plays in hip flexion by working synergistically with the hip musculature.

Clinical Relevance: The importance of the contribution of the IFL to the hip flexors warrants careful handling and repair of these ligaments in cases of surgery and structural damage.

Keywords: iliofemoral ligament; hip arthroscopy; biomechanics; anterior hip capsule; capsular repair

Bipedal locomotion has been without a doubt the most challenging mechanical adaptation in human evolution. Our ancestors, the Neanderthal and early *Homo sapiens*, were nomads. They needed to cover large distances in an energy-efficient manner to survive. Part of this mechanical efficiency originates from alternating energy transfer between gravitational-potential energy and kinetic energy within each stride. Cavagna et al[6] estimated this transfer to be greatest at intermediate walking speeds, around 3.8 km/h, accounting for up to 70% of the total energy, leaving only 30% to be actively supplied by muscles. In addition, the elastic deformation of passive structures, such as skin, ligaments, tendons, and joint capsules, may store and return energy during gait in order to accelerate the body segments.[22]

Koussou et al[22] presented a review on the contribution of passive moments to intersegmental moments during gait and reported that passive mechanisms contribute substantially to the normal human gait and that passive structures, acting as elastic springs, help reduce the energy cost of gait.[22] At the hip, the passive structures store energy at extension and subsequently deliver 20% to 50% of the moment before and during the initial swing.[22,34,39] The theory of passive energy storage for minimizing the energetic cost of locomotion is well known in the growing research domain of assistive devices and exoskeletons.[8,9] Recent work investigating passive energy storage by external

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springlike orthoses has observed up to a 23% reduction in the participants' contribution to hip power.[16]

Not surprisingly, the strongest elastic structure of the human body can be found in the iliofemoral ligament (IFL). This Y-shaped ligament, originally described by American surgeon Henry Jacob Bigelow, arises from the anterior inferior iliac spine and acetabular rim and spreads downward to the intertrochanteric line.[37] It blends with the anterior hip capsule on its deep surface and holds an ultimate tensile stress exceeding 10 N/mm.[2,32] Further, it is the most commonly incised ligament to gain access to the hip joint. The literature is inconsistent on the importance of the ligament, and there is ongoing debate on whether or not capsular repair is mandatory after capsulotomy.[4,7,13,14]

Despite the impressive mechanical and elastic properties of the IFL, literature has nearly exclusively focused on the passive, joint-stabilizing properties of the ligament.[10,18,19,24] A growing body of evidence seems to support repair, mainly in prevention of so-called microinstability, and both improved outcome and accelerated return to full activity have been demonstrated after adequate capsular repair.[11,14,21,26-28,38,41] Yet, ligaments are adaptive biological tissues. They cellularly respond and structurally adapt to their mechanical requirements.[15] If the IFL only served as an emergency break for the occasional hip hyperextension trauma and was not continuously challenged mechanically, would it be that impressively strong? Recent mechanical studies have questioned and even rejected the hypothesis that the hip capsule ligaments are solely responsible for mechanical stability of the hip.[29,32]

This brings us to the aim of the present study: defining the contribution of the IFL as an elastic and passive energy–storing assistant in human bipedal walking. By simulating walking with and without the IFL in a musculoskeletal modeling environment, we aimed to study the effects of the IFL on the required muscle work of the hip musculature.

METHODS

Study Population

For the purpose of the present analysis, straight-level walking data from 50 adults were obtained from the comprehensive gait database of Schreiber and Moissenet.[33] This study population consisted of 26 men and 24 women who were asymptomatic and had no lower or upper extremity surgery in the previous 2 years.

Detailed participant-specific information such as age (19-67 years), length (1.55-1.92 m), leg length (0.720-0.920 m), mass (50.0-98.0 kg), and body mass index (17.2-29.2) was provided, and the database described motion-capture trajectories and force-plate data at different walking speeds. These data were imported into the AnyBody Modelling Sytem (AMS)and used for participant-specific analysis. For each participant, 3 trials with a walking speed between 0.8 and 1.2 m/s were selected. An illustration of a walking participant with the inclusion of the reflective markers is shown in Figure 1.

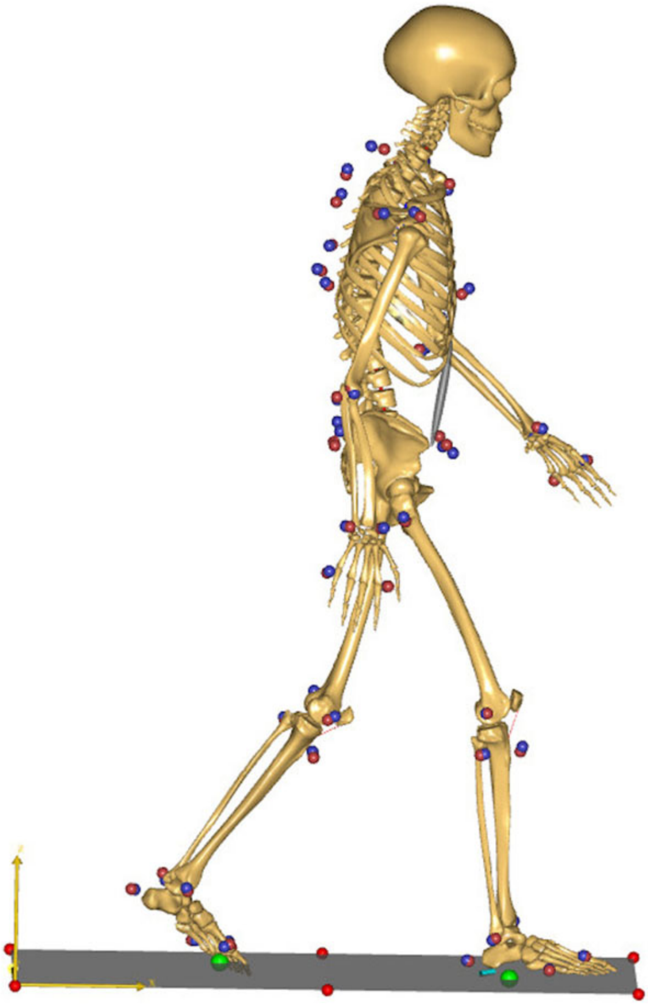

Figure 1. A snapshot of the kinematic simulation of a walking participant with locations of the reflective markers.

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Ethical approval was not sought for the present study.

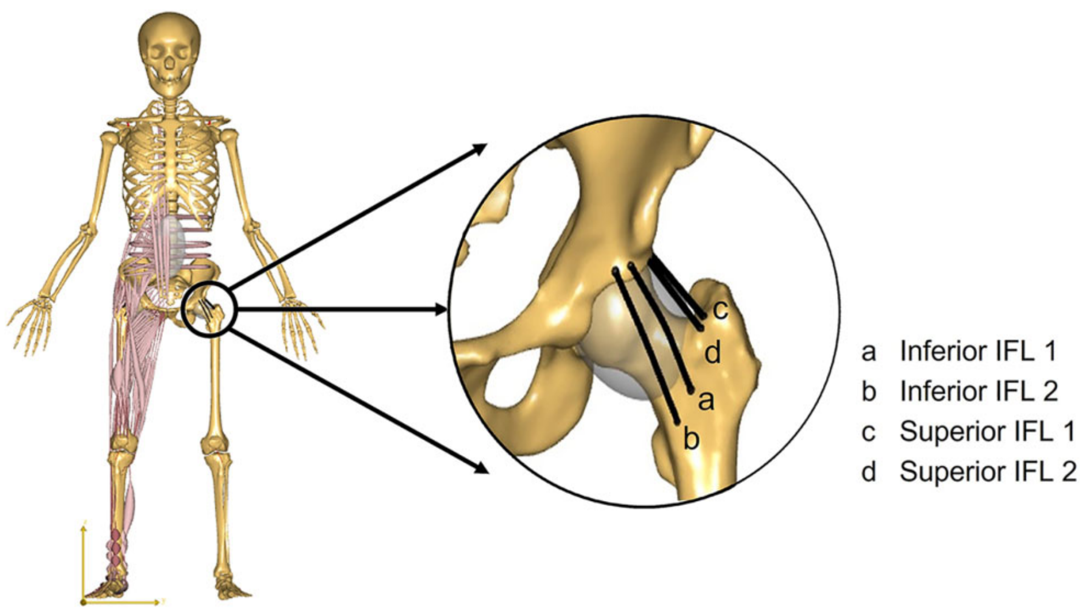

Figure 2. An illustration of the musculoskeletal model in the standing trial position. (Inset) The locations of the springs modeling the inferior (*a* and *b*) and superior (*c* and *d*) iliofemoral ligament (IFL) are indicated. The wrapping surface used for the IFL is shaded gray.

Musculoskeletal Simulation

The contribution of the IFL in the amount of muscle work required for regular walking was simulated using the AnyBody Modeling System (AMS) (Version 7.3; AnyBody Technology) with the use of the AnyBody Managed Model Repository (Version 2.3.3) and the Twente lower extremity model Version 2 (TLEM 2.1; data set[5]). We retained the standard topology of the AnyBody human model, which assumes the hip joint is a ball-and-socket joint with 3 degrees of freedom and the knee is a hinge joint with 1 degree of freedom (Figure 2). For each participant, the individual segments were nonuniformly scaled before participant-specific kinematics were imposed. For the inverse dynamic analysis, the 3-element Hill-type muscle model was used in the lower body and trunk.[20] No muscles were included for the arms. The muscle strengths were adjusted using the standard length, mass, and fat scaling law.[30] The fiber and tendon lengths of the muscles were calibrated using the proposed model of Winby et al,[40] implemented in the AMS.[17] This method calibrates the tendon and fiber lengths of a muscle at joint postures where the muscle is presumed to be at its neutral position.

In musculoskeletal modeling, the accuracy of the forces transmitted by muscles and ligaments strongly depends on the positions of these structures relative to the center of the joints, as these positions directly influence the strain, the moment arm, and the torque.[2] While wrapping can be performed up to a perfect anatomical accuracy using the original surfaces of the femur and pelvic bone, these surfaces cannot be included into musculoskeletal simulation software because of the high computational cost.[3] The wrapping objects are therefore typically replaced by geometrical wrapping obstacles to mimic the underlying anatomy.[1] To this end, we preliminarily compared ligament strains using both spherical and ellipsoid obstacles against the original femoral anatomy in a dedicated MATLAB (MathWorks) environment. An offset of 4 mm was taken into account to compensate for cartilage and ligament thickness. While in general all tests resulted in similar strain calculations, the ellipsoid obstacle mimicked the wrapping of the IFL over the acetabular rim more closely than the simple spherical approximation and was therefore selected to be included in the AMS. The superior and inferior parts of the IFL were each represented by 2 springs. The origin and insertion locations were based on the work of Tsutsumi et al[37] (Figure 2).

Kinematic Analysis

For both the inferior and the superior IFL, a linear force-strain relation was defined. The force at or below the ligament rest length was presumed to be zero. The engineering strain throughout each gait cycle was calculated by using following formula:

$$\epsilon(t) = \frac{L(t) - L_0}{L_0}$$

where $\epsilon(t)$ is the engineering strain at time point t, $L(t)$ is the length of the IFL at time point t, and L_0 is the rest length of the ligament as obtained from the calibration.

Table 1 provides an overview of the mechanical properties of the inferior and superior IFL found in the literature. We opted to apply the maximal force and maximal

TABLE 1
Mechanical Properties of the Iliofemoral Ligament Provided in Previous Studies[a]

	Hewitt et al[19]			
	Inferior	Superior	Schleifenbaum et al[32]	Pieroh et al[29]
Failure stress, MPa	6.2 ± 8.8	2.7 ± 1.4	10.0 ± 7.6	—
Maximal force, N	351.3 ± 159.4	320.3 ± 267.7	—	—
(Failure) engineering strain, %	10.4 ± 4.7	6.2 ± 1.8	84.5 ± 36.0	129.8 ± 11.1
Elastic modulus, MPa	243.2 ± 419.9	113.3 ± 63.6	24.4 ± 21.0	48.8 ± 21.4
Cross section, mm^2	92 ± 48	120 ± 40	—	53.5 ± 15.1

[a]Data are reported as mean ± SD. Dashes indicate no data available.

TABLE 2
Mechanical Properties Used to Define the Linear Force Deformation Relationship for the Parameter Study on the Inferior and Superior IFL[a]

	Adjusted Force, N		Adjusted Engineering Strain, %	
% of IFL Maximal Force[b]	Inferior IFL	Superior IFL	Inferior IFL	Superior IFL
25	87.8	80.1	10.3	7.7
50	175.7	160.15	10.3	7.7
75	263.5	240.2	10.3	7.7
100	351.3	320.3	10.3	7.7
125	439.1	400.4	10.3	7.7

[a]IFL, iliofemoral ligament.
[b]From Hewitt et al[19] (inferior IFL, 351.3 N; superior IFL, 320.3 N).

engineering strain found by Hewitt et al,[19] as they the provided maximal force and engineering strain in their well-documented analysis, whereas the other authors solely provided stress without the corresponding area.

To account for the large uncertainty involving the properties of the mechanical ligament, we conducted a parameter study. In this parameter study, the strength of the inferior and superior IFL was reduced to 25%, 50%, and 75% and increased to 125% of the maximal force as given by Hewitt et al.[19] A detailed overview of the ligament properties used in this study is provided in Table 2. For lack of a standardized methodology to personalize ligament properties, ligamentous forces were scaled proportionally using the AMS model strength scaling of the pelvic muscles. In a study on 20 healthy young adults, Silder et al[34] showed that positive passive hip flexor moments start to appear from about –5° of hip extension. Therefore, we calibrated the ligament rest length using the standing trials provided by Schreiber and Moissenet.[33]

The kinematic analysis of the participant-specific models was performed using an overdetermined kinematic solver. Next, the resultant set of joint angle trajectories was used to drive the inverse dynamics. The muscles were recruited using a second-degree polynomial muscle recruitment optimization algorithm.[7] This process allowed us to calculate the mechanical power of muscle tendon units. The mechanical work performed on the skeleton by each muscle/muscle group was calculated by integrating the absolute value of the obtained mechanical power over time.

Statistical Analysis

To test whether there was a statistical difference between the required mechanical work with and without the IFL, we calculated the mean mechanical work for each participant for each strength case of the parameter study. Next, analysis of variance was performed on 6 groups (namely, without IFL and with IFL, with strength cases of 25%, 50%, 75%, 100%, and 125%) for each separate muscle group. Subsequently, post hoc paired *t* tests were performed with Bonferroni correction. When the assumption of homogeneity of variance was violated, 1-way analysis of variance was performed with Welch correction. The Games-Howell test was then used post hoc. The statistical tests were performed in SPSS version 27 (BMI, Chicago, IL, USA) with an alpha level of 0.01.

To visualize the effect of the IFL, we calculated the ratio between the mechanical work performed by different muscles/muscle groups with and without IFL.

RESULTS

Incorporating the IFL in the musculoskeletal model did not significantly alter the required muscle work from the glutei and the quadriceps, with the exception of a minor increase in quadriceps work for the 125% strength case. It did, however, significantly influence the required work from the iliopsoas and sartorius ($P < .01$). With the IFL undamaged and fully operational, the effort required by the hip flexor muscles was reduced by medians of 54% (99% CI, 45%-62%) and 41% (99% CI, 27%-54%) for the iliopsoas and the sartorius muscles, respectively. A detailed overview of the findings of the parameter study describing the effect of the IFL at different strengths is given in Table 3. In Figure 3, these ratios are plotted using several box plots for the different strength cases for the 4 muscles/muscle groups under investigation.

TABLE 3
The Ratio of the Work Performed by the Iliopsoas and Sartorius With and Without the IFL Included in the Musculoskeletal Simulation[a]

Ratio	Iliopsoas	Sartorius
25% IFL to no IFL	0.663 (0.577-0.740)[b]	0.957 (0.899-0.987)
50% IFL to no IFL	0.611 (0.509-0.682)[b]	0.840 (0.743-0.921)
75% IFL to no IFL	0.559 (0.460-0.643)[b]	0.692 (0.610-0.822)[b]
100% IFL to no IFL	0.458 (0.376-0.547)[b]	0.592 (0.460-0.727)[b]
125% IFL to no IFL	0.490 (0.404-0.587)[b]	0.495 (0.322-0.635)[b]

[a]Data are reported as median (99% CI). IFL, iliofemoral ligament.

[b]Statistically significant difference in work performed versus the case without contribution of the IFL ($P < .01$).

An overview of the maximum obtained engineering strains during gait according to the 4 springs representing the modeled IFL is given in Table 4. The medians and 95% CIs were very similar for the 4 separate parts. The maximum engineering strain for the whole IFL is 18.7% (95% CI: 8.0%-26.5%).

DISCUSSION

There is general consensus that the ligaments and the capsule of the hip play a crucial role in stabilizing the joint.[10,24,29] However, the given study indicated that the role of these ligaments may exceed stabilization. Based on the present findings, we suggest that the IFL works synergistically with the hip musculature and thereby plays a role in energy-efficient ambulation. In absence of the IFL, the required work of the hip flexors was doubled as compared with the healthy situation.

Although the hip capsule plays an important role in the native hip function, capsular closure remains a

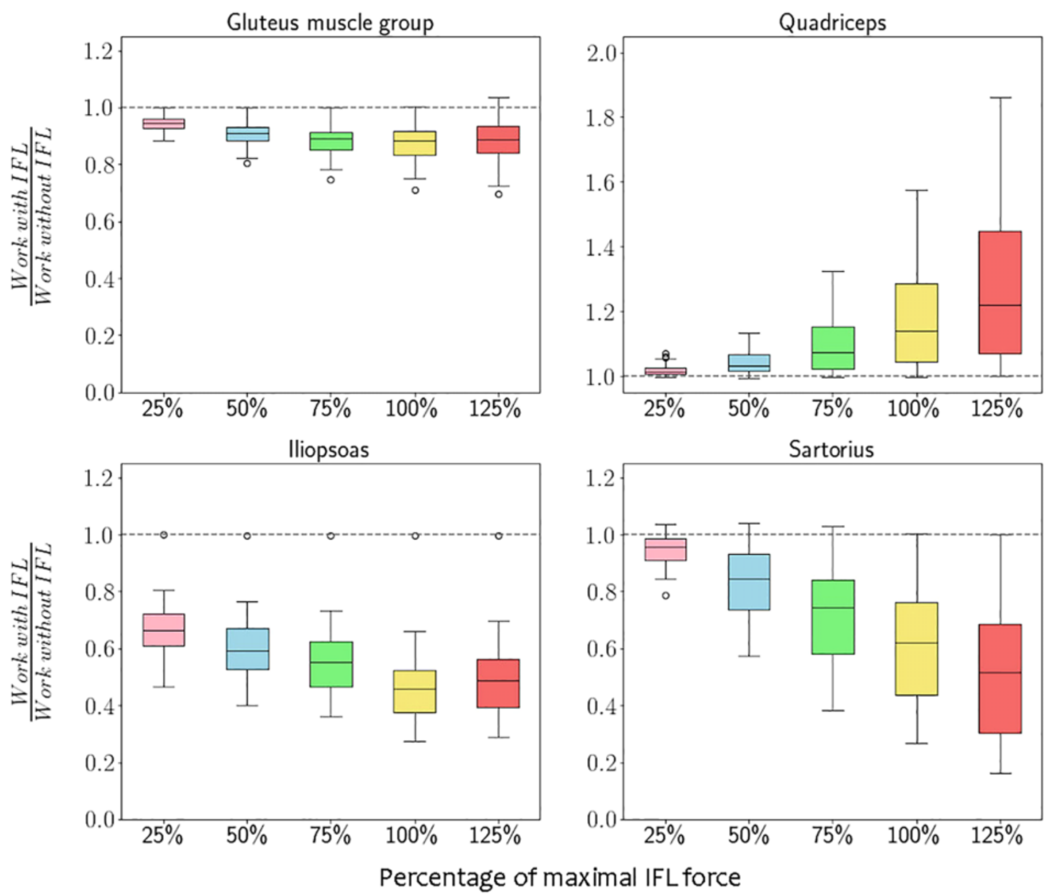

Figure 3. The ratio of work performed with and without iliofemoral ligament (IFL) inclusion for different IFL strengths (25%, 50%, 75%, 100%, and 125% of the maximal force according to Hewitt et al[19]) for the gluteus muscle group (upper left), the quadriceps (upper right), the iliopsoas (lower left), and the sartorius (lower right). The horizontal dotted line indicates where the required muscle work with and without the IFL does not differ. The circles indicate the outliers.

TABLE 4
Maximum Engineering Strain for the Different Parts of the IFL[a]

	Engineering Strain, %
Inferior IFL 1	18.8 (8.7-24.8)
Inferior IFL 2	18.0 (7.0-24.9)
Superior IFL 1	18.4 (8.6-25.9)
Superior IFL 2	21.0 (11.0-27.9)

[a]Data are reported as median (95% CI). IFL, iliofemoral ligament.

controversial topic in hip arthroscopy.[12,14,27] Previously, mainly the stabilizing function of the (anterior) hip capsule was studied.[10,18,19,24] For example, several case reports noted that without restoration of the native anatomy, the hip joint may translate during athletic activities, leading to recurrent microstability or even dislocation.[10,14,21,26] In addition, biomechanical evidence strongly supports the role of capsular repair in maintaining stability of the hip.[10,21,26,27] Further, capsular closure has also improved patient-reported outcomes. For patients treated for femoroacetabular impingement with hip arthroscopic surgery with T-capsulotomy, Frank et al[14] demonstrated superior sport-specific outcomes at 6, 12, and 24 months postsurgery for patients with a complete repaired capsule compared with patients with a partially repaired (closed vertical incision, open interportal incision) capsule at the time of surgery. More recently, Economopoulos et al[11] performed a prospective randomized trial to assess T-capsulotomy without closure (TC), interportal capsulotomy without closure (IC), and interportal capsulotomy with closure (CC). At 3, 6, 12, and 24 months after surgery, the CC group showed significantly better Hip Outcome Score–Activities of Daily Living (HOS-ADL) and Hip Outcome Score–Sports-Specific Subscale (HOS-SSS) scores compared with the TC group. At 3 months postoperatively, the HOS-ADL and HOS-SSS scores of the CC group were found to be significantly better than those of the IC group. At later time points, no significant differences between the CC and IC groups were found. These results suggest that repair after capsulotomy may be favorable and lead to quicker recovery.[11]

Our study provided a different perspective on the anterior hip capsule, more specifically on the IFL. We did not treat the IFL as a static stabilizer but rather as a part of the overall kinematic chain. Ligaments behave like elastic springs.[22] When extended, a force counteracting the extending force is generated within the spring, and potential energy is stored within the spring. Upon release or when the extending force is decreased, the potential energy is released as well. Therefore, the ligaments not only are stabilizers but also serve as energy storage systems.[22] The inferior IFL and iliopsoas share the same moment arm around the femoral head. Hence, the hypothesis was that if the inferior IFL is included in a musculoskeletal model, less work is required from the hip flexors. This study confirmed the hypothesis. Adding the IFL reduced the amount of executed work by the iliopsoas by half. A median of 41% reduction was found for the other hip flexor, the sartorius. However, the work reduction of the sartorius was characterized by more variation than for the iliopsoas. The effect on the glutei and the quadriceps was limited. Although when the strength of the IFL was increased to 125% of the mean strength, more work was required from the quadriceps. This was caused by a redistribution of the required muscle forces.

The findings provide an additional argument for meticulous handling and repair of the anterior hip capsule in the ongoing debate on capsular repair and are of particular interest for surgical procedures and approaches damaging the structural integrity of the anterior hip capsule and ligaments.

To deal with the uncertainty of the mechanical properties, we included a parameter study. We adopted the mechanical properties found by Hewitt et al[19] and reduced the strength to 25%, 50%, and 75%. The reduction in work for the iliopsoas remained significant for these 3 cases. Yet, the obtained engineering strains were larger than the failure engineering strains obtained by Hewitt et al[19] (ie, inferior IFL, 10.3% ± 5.0% and superior IFL, 7.7% ± 2.2%), and thus the obtained forces for the nonreduced strengths were higher than the failure forces. However, Myers et al[23] found a maximal force of 200 ± 100 N at maximal extension in their finite element model, which corresponds to the 25% and 50% strength cases in our study. This showed that although the obtained engineering strains were larger than those in the cadaveric experiments obtained in Hewitt et al,[19] the forces created in the IFL remained compatible with the existing literature for the 25% and 50% strength cases.[35]

A considerable strength of the present study was the inclusion of the large public gait data set of Schreiber and Moissenet,[33] which allowed covering several types of walking. Additionally, the inclusion of the parameter study allowed us to investigate the effect of the strengths of the IFL. There were 3 methodological limitations in this research that could be addressed in future research. First, only linear strength deformation relations were used for the IFL. In a typical force/displacement curve of a ligament, 3 distinct regions can be found: the toe, linear, and yield and failure regions.[31] In normal activity, most ligaments exist in the toe and linear regions.[31] The former is responsible for the nonlinear behavior in the early stage of elongation (before the length of the ligament reaches the rest length). In this region, the force counteracting the applied deformation is rather low. Therefore, we assumed that the error due to not including the toe region was limited.

A second limitation was the substantial variability in the mechanical properties of the IFL as reported in the literature. Direct comparisons among studies are difficult given the number of methodical differences, including the preservation methods of the cadaver/tissue, the age of the participants, the preconditioning of the tissue, the experimental setups, and the manner of reporting.[19,29,32,35] For example, while Hewitt et al[19] performed mechanical testing on bone-ligament-bone structures, Schleifenbaum et al[32] and Pieroh et al[29] determined the mechanical properties on ligament sections. To address this, we adopted several measures. First, we used the mechanical properties that were previously utilized by other authors[42] and/or that corresponded to measurements of others.[23,35,36] In addition, a parameter study was included. Further, the rest length of the ligament was determined based on the standing trials of the participants to ensure that the natural pelvic tilt was taken into account.

A third limitation was that no participant-specific bone morphology was used. Instead, for each participant the generic bone morphology was nonuniformly scaled using standard scaling laws implemented in the AMS.[30] The wrapping objects were scaled accordingly. Using scaled generic bones and wrapping objects introduced an idealization of geometry, and although this implied that for some participants the moment arm of the iliopsoas and IFL was possibly inaccurate, we considered this to be acceptable for this proof-of-concept study.

CONCLUSION

The anterior hip capsule assists the hip flexors during normal ambulation. Meticulous handling and repair of the structure seems important in any procedure affecting its integrity. Clearly, further investigations are required to determine the in vivo mechanical properties of the IFLs. Furthermore, the effect of surgical repair and scarring[25] on the passive mechanical and elastic properties of the capsule and ligaments and on passive assistance during ambulation needs to be investigated.

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