


Article

Intramuscular Properties of Resting Lumbar Muscles in Patients with Unilateral Lower Limb Amputation

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Featured Application: This study can assist in forming a basis for the therapeutic strategies for the prevention and treatment of low back pain in amputees.

Abstract: Lower limb amputees (LLAs) have a high incidence of low back pain (LBP), and identifying the potential risk factors in this group is key for LBP prevention. This study analyzed the intramuscular properties of the resting lumbar muscle in thirteen unilateral LLAs and age-matched controls to predict the onset of LBP. To measure the lumbar intramuscular properties, resting erector spinae muscles located in the upper and lower lumbar regions were examined using a handheld myotonometer. The dynamic stiffness, oscillation frequency, and logarithmic decrement were measured. In our results, the stiffness and frequency of the upper lumbar region were greater in the amputee group than in the control, whereas the decrement did not differ between the two groups. Additionally, the measured values in the lower lumbar region showed no significant difference. Within each group, all three factors increased at the upper lumbar region. In the LLAs, the frequency and stiffness values of the upper lumbar on the non-amputated side were significantly higher than those on the amputated side. These results indicate that the upper lumbar muscles of the amputees were less flexible than that of the control. This study can help in providing therapeutic strategies treating LBP in amputees.

Keywords: lower limb amputation; low back pain; lumbar intramuscular properties; myotonometry

Citation: Chang, Y.; Kang, J.; Kim, G.; Shin, H.; Park, S. Intramuscular Properties of Resting Lumbar Muscles in Patients with Unilateral Lower Limb Amputation. *Appl. Sci.* **2021**, *11*, 9122. <https://doi.org/10.3390/app11199122>

Academic Editor: Joung Hwan Mun

Received: 2 September 2021

Accepted: 28 September 2021

Published: 30 September 2021

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1. Introduction

Low back pain (LBP) is a common musculoskeletal disorder among lower limb amputees (LLAs), where 52% of LLAs experienced persistent, severe LBP and had a greater prevalence of LBP (52–89%) than that of the non-amputees [1–3]. Identification of potential risk factors in people at high risk for LBP is critical for disease prevention and management.

Abnormal lumbar muscle activity is one of the factors affecting the LBP of LLAs. Factors contributing to left-right asymmetry include static asymmetry (i.e., leg length discrepancy, LLD) and dynamic asymmetry (i.e., asymmetrical trunk-pelvic movement) (Figure 1) [4–6].

With respect to static asymmetry, LLAs generally use prosthetic legs for walking after amputation, which are intentionally shortened to facilitate the toe clearance of the prosthesis. Many previous studies have reported the use of short prosthetic legs by LLAs [4,7], and that a LLD greater than 30 mm can affect the lumbo-pelvic kinematics [8].

When the length of the prosthesis is shortened, the pelvic inclination on the prosthetic side is lowered [8], and when exposed to this static asymmetry for a long time, the prosthetic pelvic muscle is stretched and the contralateral lumbar muscle is shortened. This characteristic causes asymmetry of the lumbar muscles, resulting in structural problems such as changes in the curvature of the spine (i.e., scoliosis) [9–11].

Dynamic asymmetry is also closely related to the amputee's gait, as LLAs excessively recruit the trunk to advance the prosthesis when walking. These characteristics increase

the lateral flexion and axial rotation of the trunk towards the prosthetic leg and increase the activity of the global trunk muscle [5]. Excessive muscle activity and trunk-pelvis motion during walking increase the spinal load, and exposure to repetitive walking increases the risk of LBP [12]. A previous study reported that the trunk muscle force and spinal load in the non-amputated side (NAS) increased by 10–40% and 17–95%, respectively, compared with those in the amputated side (AS), which changed the recruitment pattern of trunk muscles [13].

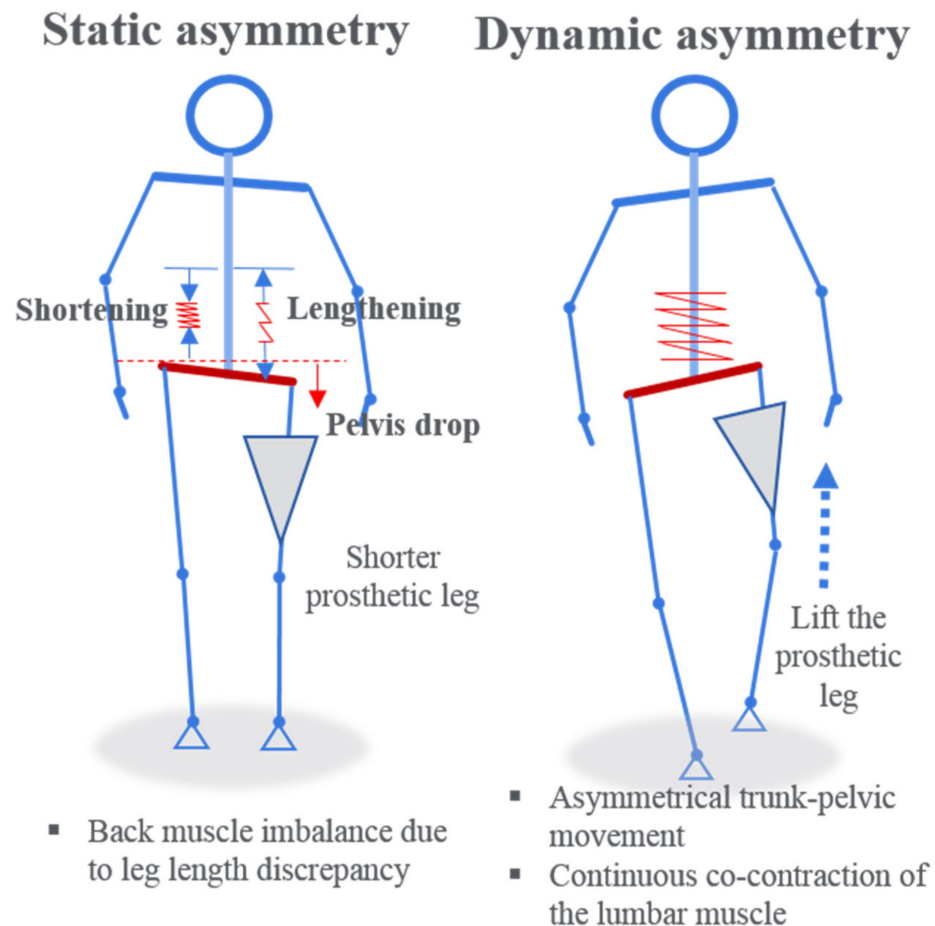


Figure 1. Schematic of static and dynamic contributing factors associated with lumbar muscle imbalance in patients with unilateral transfemoral amputation.

Lumbar muscle activity patterns during LLA's walking are similar to those in patients with LBP [14,15]. In healthy individuals, the erector spinae muscle relaxes during the double limb support and swing phases, but in patients with LBP, the activity of the erector spinae muscle is increased throughout the entire gait cycle [15–17]. The activity of erector spinae muscle in patients with chronic low back pain (CLBP) is 8 to 48% higher than in healthy individuals [14]. This is closely related to the guarding mechanisms that protect the spine from pain.

However, LLAs increase the activity of the erector spinae as a functional strategy for gait propulsion, not due to LBP [12]. In preparing the trunk for movement of the prosthetic leg, it is necessary to provide stability to the lumbar region, which increases the muscle activity of the trunk. As a result, the activity of the erector spinae muscle increases throughout the gait cycle, and the recruitment pattern of the trunk muscles may change, similar to that in patients with LBP.

As mentioned above, the asymmetry of pelvic inclination caused by LLD or trunk-pelvis asymmetry movement due to abnormal gait changes the onset or recruitment pattern of the back muscles, causing overall lumbar muscle imbalance. Abnormal patterns

or excessive activation of muscles can lead to musculoskeletal problems in the long term. Therefore, it is of utmost importance to identify potential risk factors to prevent them.

Many studies have been conducted to analyze the properties of muscles in relation to LBP. In particular, analyzing the properties of resting lumbar muscles is useful to identify potential risk factors for back pain [18], and analyzing the stiffness of resting lumbar muscles enables the analysis of inherent skeletal muscle properties in a state where central nervous system (CNS) activation is excluded [19].

To identify the lumbar muscle properties, the stiffness is mainly analyzed. High muscle stiffness means high muscle tone, and prolonged exposure to these situations is interpreted as a high risk of muscle fatigue or LBP [20].

In previous studies, authors reported increased lumbar muscle tone in patients with CLBP and a 20% greater stiffness of the erector spinae muscles at rest, as compared to that in patients without LBP [21]. High stiffness of the spinal muscles indicates hypertonia due to chronic mechanical overload with or without an inflammatory response [22]. A high muscle stiffness value is a valid factor for predicting the occurrence of LBP, and relevant evidence has mainly been studied in patients with ankylosing spondylitis [23]. However, studies in LLAs with high prevalence of LBP are rare.

Intramuscular properties are measured using a variety of methods such as diagnostic ultrasound, magnetic resonance elastography, ultrasonic shear-wave elastography, and electromyography [24]. Although these methods can provide high reliability and quantified data, their high cost and operational complexity can make them difficult to use in simple clinical practice.

Several recent studies have investigated intramuscular properties using a handheld myotonometer. It is a non-invasive device that can quantify stiffness, frequency, and decrement of myofascial tissue, and real-time quantitative measurement of myofascial stiffness can support diagnostics and patient-driven interventions. Studies using a myotonometer were conducted in patients with ankylosing spondylitis [23,25], Parkinson's disease [26], and stroke [27], and the reliability was confirmed with an intraclass correlation coefficient of 0.80 or greater [28].

As mentioned above, most previous studies related to LBP in LLAs have focused on the kinematic changes such as trunk and lumbo-pelvic movement asymmetry [5,6] as well as variations in the trunk-pelvis coordination during walking [14]. Although some studies have reported asymmetry of lumbar muscle activity using surface electromyography [29], to the best of the authors' knowledge, there are few studies on the changes in the intramuscular properties of lumbar muscles in LLAs. Understanding the biomechanical characteristics (i.e., the intramuscular properties) of groups with a high prevalence of LBP will help prevent disease and provide adequate treatment by identifying the potential risk factors.

Therefore, this study aims to compare the biomechanical properties of lumbar muscles using a handheld myotonometer in LLAs and healthy controls. The results of this study will be utilized as basic data for establishing rehabilitation strategies for amputees with LBP.

2. Methods

2.1. Participants

Thirteen LLAs and thirteen age matched controls participated in this study. The general characteristics of participants are shown in Table 1.

Table 1. General characteristics of the participants ($n = 26$).

Amputated Level (n)	Amputee Group ($n = 13$)	Control Group ($n = 13$)	t-Value
		TFA (10)/TTA (3)	
Amputated side (n)	Right (5)/Left (8)	NA	NA
Prosthesis use (years)	4.5 ± 1.1 *	NA	NA
Prosthesis use time per day (hours)	9.2 ± 1.8	NA	NA
Age (years)	43.2 ± 9.9	42.5 ± 5.5	0.221
Weight (kg)	70.5 ± 6.2	73.1 ± 4.7	-1.196
Height (m)	1.71 ± 4.13	1.75 ± 0.06	-1.800
BMI (kg/m^2)	23.6 ± 2.1	24.0 ± 1.9	-0.581

* Value is presented as mean \pm standard deviation; TFA, transfemoral amputation; TTA, transtibial amputation; BMI, body mass index; NA, not applicable.

As a result of verifying the subjects' homogeneity using the independent sample t -test, there was no statistically significant difference between the groups in age, weight, height, and BMI ($p > 0.05$).

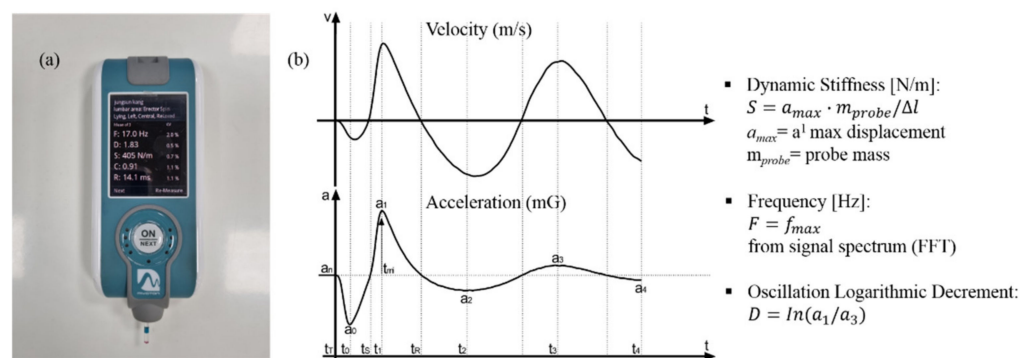
The inclusion criteria for amputees were as follows: male patients with unilateral transfemoral and transtibial amputation due to post traumatic injury; patients with no history of neurological symptoms except phantom and stump pains; patients without problems in daily activities (i.e., walking) and ability to walk without assistive devices; and patients without medication or physical therapy due to LBP within 3 months. In addition, the control group (CG) was selected from healthy individuals without any medications and history of LBP in the past 3 months.

In particular, those with pre-existing spinal pathology or chronic LBP prior to traumatic amputation, coexistence of spinal trauma that occurred at the time of traumatic injury, those with persistent LBP for 3 months or more due to the possibility of fatty degeneration of the spinal muscles [30], and those with a BMI greater than $30 \text{ kg}/\text{m}^2$ were excluded from this study to improve the reliability of quantification of intramuscular properties [31].

The purpose of the study was explained to all participants and informed consent was obtained. The study was conducted with the approval of the Research Ethics Committee of the Rehabilitation Engineering Research Institute (RERI-IRB-190114-1).

2.2. Instruments

Lumbar intramuscular properties were measured using a MyotonPro[®] (Myoton AS, Tallinn, Estonia); it is an objective and reliable non-invasive measurement device that can quantitatively evaluate the biomechanical properties of soft tissues such as muscles and tendons [23,28]. Figure 2 shows the actual experimental status and extracted data using MyotonPro[®].

**Figure 2.** Myotonometer used in the experiment (a) and the extracted signal characteristics (b).

The MyotonPro[®] provides a controlled pre-load of 0.18 N for initial compression of the subcutaneous tissue and imposes an additional 15 ms impulse of mechanical force of 0.40 N to induce a damped or decaying natural oscillation of the tissue [32]. The peak acceleration (α_{\max}) of the natural oscillation is estimated using an accelerometer. The first integral of the acceleration signal extracts velocity and the second integral extracts the displacement. The MyotonPro[®] quantifies dynamic stiffness (S), as defined in Equation (1)

$$S = \frac{m\alpha_{\max}}{\Delta l} \quad (1)$$

where m is the mass of the probe of 18 g (0.18 N), α_{\max} is the maximum amplitude of the oscillation in the acceleration signal, and Δl is the amplitude of the displacement signal after the end of the impulse time. The MyotonPro[®] can measure tissue stiffness within a subcutaneous depth of 2 cm [23].

2.3. Measured Variables

The measured variables include the dynamic stiffness (N/m), oscillation frequency (Hz), and logarithmic decrement. Dynamic stiffness (i.e., tone) is an intramuscular property defined as the resistance to deformation that occurs when external force is applied. Oscillation frequency (i.e., resting tension) refers to the intrinsic muscle tension in the absence of spontaneous muscle contraction.

Logarithmic decrement (i.e., elasticity) is measured as the damping ratio of the signal of acceleration as the tissue recovers after deformation. A decrease in the logarithmic decrement value indicates an increase in tissue elasticity and a decrease in tissue plasticity [18,25,32].

2.4. Experimental Procedure

Subjects were placed in a prone position on an examination table with the skin of the lumbar area exposed, and their arms were placed next to the torso for a comfortable posture.

To measure the lumbar intramuscular properties, we selected four sites around the spine: the upper (i.e., T12–L1) and lower lumbar regions (i.e., L4–L5) as shown in Figure 1. The upper lumbar region includes the erector spinae muscles, which corresponds to the inflection point of the thoracolumbar curvature, where high loads and rotational forces occur [33]. The lower lumbar region contains the multifidus muscles, which play a key role in maintaining stability of the spine [34]. Lumbar measurement levels were initially approximated in the uppermost part of the iliac crests, then the specific levels were identified the L4–L5 and T12–L1 spinous processes, and bilateral measurement sites were marked on the left and right extensor muscle bulk prominences (Figure 3) [18].

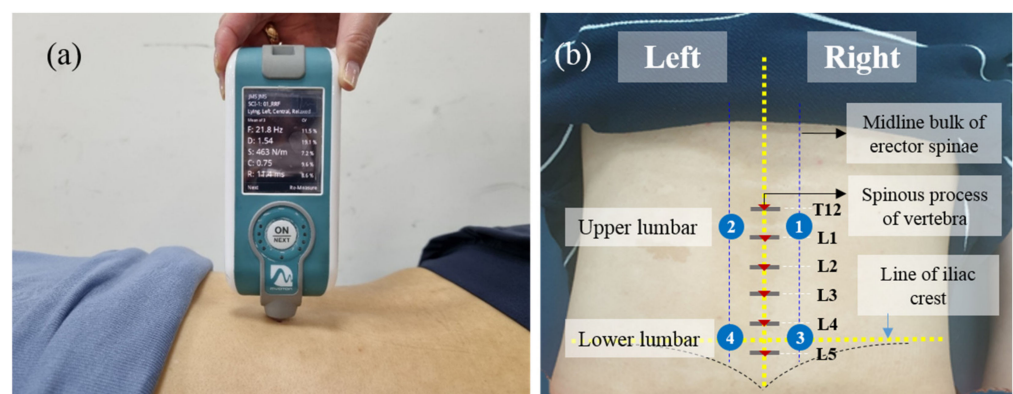


Figure 3. A photograph of the measurement using a myotonometer (a) and a schematic of the selection criteria for the four measurement sites of the lumbar (b).

Before measuring, the subjects held a relaxed position without muscle contraction for 10 min. The stiffness of the resting lumbar fascia is known to increase after 10 min from its initial state and is also a fundamental intrinsic property of the fascia [18]. After rest, the lumbar intramuscular properties of the subjects were measured while they held their breath for 5 s at the end of the inspiratory phase to minimize the effect of the intra-abdominal pressure [35]. All MyotonPro[®] measurements were conducted by the same skilled physical therapist to minimize measurement errors.

2.5. Statistical Analysis

Statistical analysis was performed using the Statistical Package for the Social Sciences, Statistics version 20 (IBM Corp., Armonk, NY, USA). The mean and standard deviation of all data were calculated, and the normality of the data was verified using the Kolmogorov–Smirnov test.

Within each group, the differences in the intramuscular properties between the right and left of the spine and between the upper and lower lumbar regions were compared using a paired *t*-test. In addition, the difference between the two groups in the three conditions (i.e., the AS, NAS, and CG) were compared using analysis of variance (ANOVA). Tukey's multiple comparison test was used for post-hoc analysis, and the statistical significance for all tests was set at $p < 0.05$.

3. Results

3.1. Differences in Lumbar Intramuscular Properties of the AS (or Right) and NAS (or Left)

Table 2 shows the differences in lumbar intramuscular properties of AS and NAS within the amputee group.

Table 2. Comparison of lumbar intramuscular properties between AS and NAS sides in the amputee group.

	Lumbar Part	Amputee Group ($n = 13$)		SE	t-Value
		AS	NAS		
Stiffness (N/m)	Upper	303.2 ± 51.3 §	329.1 ± 51.2	8.57	−3.026 *
	Lower	225.9 ± 51.2	237.6 ± 61.3	8.73	−1.348
	SE	16.89	17.92	-	-
	t value	4.579 ***	5.105 ***	-	-
Frequency (Hz)	Upper	16.5 ± 1.9	17.4 ± 1.5	0.30	−2.704 *
	Lower	14.0 ± 1.4	14.1 ± 1.5	0.33	−0.503
	SE	0.60	0.58	-	-
	t value	4.318 ***	5.590 ***	-	-
Decrement (Ratio)	Upper	1.25 ± 0.24	1.17 ± 0.17	0.06	1.379
	Lower	1.15 ± 0.23	1.13 ± 0.23	0.04	0.496
	SE	0.06	0.06	-	-
	t value	1.881	0.744	-	-

AS, amputated side; NAS, non-amputated side; SE, standard error; § mean ± standard deviation; * $p < 0.05$, *** $p < 0.001$.

Mean values of stiffness and frequency were greater in NAS than in the AS for both upper and lower lumbar regions; the upper and lower lumbar stiffness increased by 8.6% and 5.2%, and the upper and lower lumbar frequency increased by 5% and 1.2%, respectively (statistical significance was only in the upper lumbar). The decrement values were less in the NAS than in the AS in both upper and lower lumbar regions (decreased by 6.2% and 1.6%, respectively); however, there was no significant difference.

Table 3 shows the differences in intramuscular properties of the right and left sides within the CG.

Table 3. Comparison of lumbar intramuscular properties between the right and left sides in the control group.

	Lumbar Part	Control Group (<i>n</i> = 13)		SE	t Value
		Right	Left		
Stiffness (N/m)	Upper	282.1 ± 38.7 [§]	281.6 ± 39.3	5.14	0.169
	Lower	225.6 ± 47.1	227.2 ± 41.2	3.02	0.130
	SE	8.55	9.68	-	-
	t value	6.375 ***	5.580 ***	-	-
Frequency (Hz)	Upper	15.3 ± 1.3	15.8 ± 1.3	0.29	−1.807
	Lower	14.0 ± 1.5	14.3 ± 1.6	0.11	−2.432 *
	SE	0.39	0.41	-	-
	t value	3.124 **	3.023 *	-	-
Decrement (Ratio)	Upper	1.29 ± 0.34	1.13 ± 0.25	0.04	4.487 ***
	Lower	1.02 ± 0.20	1.00 ± 0.16	0.03	0.506
	SE	0.07	0.04	-	-
	t value	3.921 **	3.010 *	-	-

SE, standard error; [§] mean ± standard deviation; * *p* < 0.05, ** *p* < 0.01, *** *p* < 0.001.

The right and left stiffness values were not significantly different in either upper or lower lumbar regions. Frequency values were greater on the left side for both upper and lower lumbar regions (increased by 3.4% and 2.0%, respectively); however, statistical significance was found only in the lower lumbar region. Conversely, decrement values were greater on the right side in both upper and lower lumbar regions, with 12.4% and 1.5% increase, respectively (statistical significance was only in the upper lumbar region).

3.2. Differences in the Lumbar Intramuscular Properties of the Upper and Lower Regions

Table 2 shows the differences in the lumbar intramuscular properties of the upper and lower regions within the amputee group. Mean values of stiffness and frequency were greater in upper than in the lower lumbar region for both AS and NAS. In the upper region, the stiffness of the AS and NAS increased by 25.5% and 27.8%, and the frequency increased by 15.6% and 18.6%, respectively, all statistically significant. The decrement value was greater in the upper than in the lower lumbar region in both AS and NAS (increased by 8.3% and 3.7%, respectively); however, there was no significant difference.

Table 3 shows the differences in lumbar intramuscular properties of the upper and lower regions within the CG.

The stiffness and frequency values were greater on the upper lumbar region for both the right and left sides; in the upper region, the stiffness of the AS and NAS increased by 19.3% and 19.2%, and the frequency increased by 8% and 9.2%, respectively, all statistically significant. In addition, decrement values were greater in the upper region in both the right and left sides (21% and 11.1% increase, respectively) and there was a statistical significance.

3.3. Differences in Lumbar Intramuscular Properties between Amputees and the Control Group

Table 4 shows the differences in lumbar intramuscular properties for the three conditions (i.e., AS, NAS, CG), and the presented CG values are the mean values of the right and left sides.

Table 4. Comparison of lumbar muscle properties in AS, NAS, and CG.

	Lumbar Part	Lower Limb Amputees		CG (C)	F Value	Tukey HSD
		AS (A)	NAS (B)			
Stiffness (N/m)	Upper	303.2 ± 51.3 §	329.1 ± 51.2	281.9 ± 38.0	3.291 *	B>C
	Lower	225.9 ± 51.2	237.6 ± 61.3	226.4 ± 43.6	0.192	NS
Frequency (Hz)	Upper	16.5 ± 1.9	17.4 ± 1.5	15.5 ± 1.2	4.455 *	B>C
	Lower	14.0 ± 1.4	14.1 ± 1.5	14.2 ± 1.6	0.082	NS
Decrement (Ratio)	Upper	1.25 ± 0.24	1.17 ± 0.17	1.21 ± 0.29	0.336	NS
	Lower	1.15 ± 0.23	1.13 ± 0.23	1.01 ± 0.18	1.184	NS

AS, amputated side; NAS, non-amputated side; CG, control group; § mean ± standard deviation; * $p < 0.05$; NS, no significance.

In the upper lumbar region, the stiffness of the AS and NAS tended to be 14.4% greater than that of the CG (F value = 3.291, $p < 0.05$). In addition, the frequency of the AS and NAS tended to be 10.9% higher than that of the CG (F value = 4.455, $p < 0.05$). Especially in the NAS, there were significant differences among the three conditions. In the post-hoc analysis (Tukey's HSD), the stiffness and frequency of NAS significantly increased compared to the CG ($p < 0.05$). On the other hand, there was no statistical difference in the stiffness and frequency of the lower lumbar region.

In addition, the decrement values of the NAS were less than those of the AS or CG (decreased by 12.4% and 6.6%, respectively), but were not statistically significant.

4. Discussion

The identification of potential risk factors for LBP is crucial in terms of disease prevention and treatment. This study analyzed the properties of resting lumbar muscles using a handheld myotonometer for LLAs with a high risk of LBP to understand LBP in patients with lower limb amputation.

The most notable point in the results of this study is the significant increase in the stiffness and frequency of the upper lumbar region on the NAS of amputees. The stiffness of the NAS (329.1 N/m) increased by 8.5% and 16.9%, respectively, compared with those of the AS (303.2 N/m) and CG (281.6 N/m), and the frequency of the NAS (17.4 Hz) was increased by 5.5% and 12.3%, respectively, compared with that of the AS (16.5 Hz) and CG (15.5 Hz).

In particular, in the amputees, the difference between the AS and NAS of the upper lumbar was significantly increased more than that in the lower lumbar. However, there was no significant difference between the right and left sides of the upper lumbar in the CG. Therefore, unlike the CG, bilateral differences in stiffness and frequency can be interpreted as a potential risk factor related to LBP in the amputees.

In our study, there was no difference in the stiffness of the right and left portions in the CG. However, in a similar previous study using the portable MyotonPro, comparing the stiffness of the right and left sides of 20 young male and female adults, it was reported that males and the right side had greater stiffness than females and the left side, respectively. This was affected by the right-hand dominance (all subjects in the study had right-handed dominance) and low-level CNS activation, suggesting further study of the effect of gender difference [18]. In a study by Nair et al., the stiffness was measured a total of three times with an interval of one week, and the difference in mean values measured over 3 weeks was compared. The men who participated in the experiment showed stiffness values in the range of 231.7 N/m to 267.7 N/m, and there was little difference between the left and right mean values at week 1 (234.6 N/m vs. 236.3 N/m). However, it can be seen that the difference between the left and right increased significantly in the second (242.9 N/m vs. 257.5 N/m) and third weeks (231.7 N/m vs. 264.3 N/m). In the discussion, the authors said that the inter-individual variance increased more than the intra-individual variance

during the 3 weeks of measurement. They did not specify the reason for this, and they suggested further research. However, it is not appropriate to directly compare the results of our study with that of Nair et al. because our experiment was conducted once a day and the change over time was not considered. Nevertheless, it was confirmed that the average stiffness values presented by Nair et al. were lower than those of the CG and amputees in our study. In addition, in the study of Nair et al., the average age of male subjects was 21 years, and unlike our study, the subjects were younger, and the measurement site (L3–L4) was different. Owing to these factors, it is thought that the lumbar stiffness of the CG was higher in our study. In the future, we believe that additional research on the difference in stiffness according to age and measurement site of individuals without LBP is necessary [18].

In general, the tendency of changes in stiffness and frequency values is similar because of the characteristics of the two variables (when the stiffness value increases, the frequency value also increases). Through these factors, it is possible to evaluate the muscle tone (or tension), which is an intrinsic property of the muscle. High muscle tone means high intramuscular pressure (i.e., hypertonia), which reduces blood supply and can lead to muscle fatigue [36]. Accumulated muscle fatigue can cause back pain; hence, proper management is required.

In our study, the stiffness and frequency differences between the right and left sides of amputated patients were associated with providing an asymmetric static and dynamic environment to the amputated patients.

First, one of the factors contributing to static asymmetry in amputated patients is LLD [6]. Several previous studies involving LLD have reported that the leg lengths of the amputees are asymmetrical; 85% of amputees had prostheses that were too short or too long ($n = 113$, 29 TFAs and 84 TTAs) [4], and 14 out of 17 amputees had prostheses that were too short [7]. It has also been reported that the LLD of the amputees with LBP is greater than that of amputees without LBP, and LLDs greater than 30 mm affect the lumbo-pelvic kinematics [8]. The anterior pelvic tilt angle of TFA increased when standing or walking compared with that of the CG [37].

Intentionally shortening the length of the prosthesis to facilitate toe clearance on the prosthetic side during TFA walking causes the problem of descent of the pelvis of the prosthesis [8]. Asymmetric pelvic oblique angles result in muscle imbalance, stretching the lumbar muscles of the short leg (the AS) and shortening the lumbar muscles of the contralateral (the NAS). These characteristics can result in a prolonged distortion of the asymmetric pelvis or fixation of the imbalance in muscle condition, resulting in scoliosis [9]. In fact, functional scoliosis due to LLD occurs frequently in LLAs [10], and scoliosis has been reported in 43% of LLAs (18 out of 42) on radiological examination [11].

Second, the asymmetry of the left and right lumbar muscles in the amputees is closely related to gait. Abnormal gait patterns in amputees cause asymmetric trunk movement, which in turn alter muscle recruitment patterns.

In general, during normal gait, the thoracic erector spinae muscle on the opposite side of the stance leg contracts concentrically before the lumbar erector spinae muscle is activated. This characteristic contributes to moving the upper trunk while reversing the curve of the spine toward the swing leg. Thereafter, the lumbar erector spinae muscle controls the trunk and supports the pelvis and swing leg elevation through eccentric contraction [38,39].

When trunk neuromuscular control is compromised by LBP, the timing of activation of trunk muscle tissue is altered as above [15,17].

For example, people with LBP had increased lumbar and thoracic erector spinae activation during the swing phase of gait, early onset and delayed offset of the lumbar erector spinae, and increased co-contraction of trunk flexors and extensors compared with those without LBP [14,15,17].

This phenomenon is a mechanism to protect the spine from LBP by increasing the stability of the spine. When LBP patients walk at a high speed, increasing the lumbar

erector spinae activity during swing phase to increase the stiffness of the lumbar spine is also effective for speed-dependent perturbations [14].

Amputees also increase lateral flexion and axial rotation of the trunk toward the amputated leg when walking, and these properties increase the activity of the global trunk muscles. This is a strategy that uses the proximal (i.e., trunk) to advance the amputated limb, and excessive muscle activity and trunk-pelvic movement increase the spinal load, increasing the risk of LBP [5,13].

With regard to spinal load, Shojaei et al. found that the trunk muscle force and spinal load were 10–40% and 17–95% higher, respectively, in the stance phase (i.e., heel strike and toe-off phase) to the sound side during TFA walking compared with the AS. At the same time, increases of 6–80% and 26–60% have been reported compared to those in normal controls. This occurs because the antagonist muscles contract simultaneously during the non-amputated stance phase, and at the same time as the recruitment pattern of the trunk muscles changes, the trunk is mainly used to assist the advancement of gait. The author of this paper suggested that repeated exposure to these high loads may increase the risk of LBP given the repetitive nature of walking [13].

Moreover, in this study, decrement factors other than stiffness and frequency were analyzed. The decrement refers to the elasticity of a muscle, and as stiffness or frequency increases, the decrement decreases. Maintaining adequate elasticity enables efficient muscle activity with less mechanical energy [32], whereas a decrease in elasticity causes muscle fatigue and restricts the speed of movements [40].

As indicated by the results of this study, there were significant differences in stiffness and frequency between amputees and controls, but there was no significant difference in decrement between the two groups. On the other hand, as for the average value of decrement, the elasticity of the non-amputated upper side was the lowest, and the elasticity of the contralateral side (the AS) was higher than that of the CG. In addition, the average value of the lower lumbar decrement showed a decrease in elasticity on both sides of the amputee group compared with that in the CG.

In conclusion, there was no significant difference in the decrement factor, whereas the stiffness or frequency values of the upper lumbar in our study showed significant differences within or between groups. This means that it is difficult to identify potential risk factors for LBP using the decrement value. In addition, the fact that there was a significant difference between the left and right of the upper decrement within the CG makes it more difficult to use as an evaluation factor. This may be the reason why most previous studies have focused on stiffness to identify potential risk factors associated with back pain.

In addition, in the results of this study, significant differences in intramuscular properties were observed in the upper region of the lumbar rather than in the lower region. In both groups, the stiffness and frequency of the upper region were significantly increased compared with those in the lower region, because the muscle arrangement or properties of the upper and lower lumbar were different. This is because the upper and lower lumbar regions are composed of muscle fibers and superficial fascia, respectively [41].

As mentioned above, the erector spinae muscle measured in this study corresponds to the inflection point of the thoracolumbar curvature [42], which is a region where high loads [33] and rotational forces [43] occur even in the upper lumbar region (T12–L1). The lower lumbar region (L4–L5) contains the multifidus muscles that play an important role in maintaining spine stability. In fact, the upper stiffness value of our study tended to be higher than those of Nair et al. Although our subjects were older, the lumbar measurement level was higher than that of Nair et al. (L3–L4), which is thought to be because this area is the inflection point of the thoracolumbar curve and stress is more concentrated in this area [18].

In addition, in both groups, all three factors were significantly higher in the upper than in the lower lumbar region. The higher muscle tone and lower elasticity in the upper

region than in the lower region means that muscle fatigue is more likely to occur in the upper lumbar region.

In general, relaxation of the erector spinae muscle occurs during the double limb support phase and swing phase during walking [16]. However, in patients with CLBP, the activity of the erector spinae muscle increases throughout the gait cycle [15,17].

Insufficient muscle relaxation and sustained activity of motor units contribute to muscle damage and degeneration [44].

The pattern of muscle activation during walking in LLAs is similar to that in LBP patients. The initial activity of the lumbar erector spinae prepares the trunk for movement towards the amputated leg by providing stability to the lumbar spine before the heel of the amputated leg contacts the floor. This unique neuromuscular control strategy of LLAs translates into a functional strategy for locomotion than that of LBP patients. These characteristics support the increase in the non-amputated upper stiffness in our findings. Gait strategies that require greater activation of the trunk muscles of the supporting leg to provide stability to the lumbar region for propulsion on the AS are thought to increase muscle stiffness and frequency, especially in the upper part on the NAS [12].

Continuous activation of muscles can cause musculoskeletal problems in the long term, so it should be prevented. Nevertheless, the imperative question is which strategy to apply to prevent these musculoskeletal problems. If increased lumbar muscle activity is associated with a maladaptive coping method, treatment strategy such as myofeedback may help reduce the increased lumbar muscle activity.

As the asymmetry of the left and right muscle stiffness of the amputees shown in our study is more likely to develop into LBP if it persists for a long time, it is of utmost importance to eliminate risk factors that cause asymmetric characteristic in amputated patients.

As mentioned above, LLD identified as a static risk factor should be aligned as symmetrically as possible, and pelvic descent due to leg length should be avoided. In addition, asymmetric movements of the trunk and pelvis caused by abnormal gait patterns should be improved through prosthetic gait training. In particular, proper muscle strengthening and education are required to prevent excessive trunk rotation, extension, and lateral flexion during the training process.

Therefore, it is necessary to establish a strategy to relieve LBP by regularly evaluating the lumbar muscle condition of LLAs. Furthermore, we believe that myotonometry can be useful as a simple non-invasive measurement tool for clinical evaluation. If the stiffness of the muscle is high, the muscles should be relaxed through stretching or massage. It is also necessary to improve the exercise methods and lifestyle to strengthen the back muscles through patient education.

This study has several limitations. Although voluntary muscle contraction was restricted by taking sufficient rest before the experiment, complete restriction of muscle contraction could not be confirmed by quantitatively measuring the electrical activity of the muscles. Future studies should consider concurrent electromyography to ensure the reliability of the results. In addition, it is necessary to analyze the muscle properties according to the level of amputation or the presence or absence of LBP and the increase in the number of subjects.

5. Conclusions

Regarding the high prevalence of LBP among patients with lower limb amputation, we analyzed the biomechanical and viscoelastic properties of lumbar muscles in unilateral LLAs. We found that the frequency and stiffness values of the upper lumbar muscles in LLAs were greater than those in the CG. In particular, the frequency and stiffness values of the NAS were greater than those of the AS. The decrement values were not significantly different between the two groups, and within each group, all three factors increased in a statistically significant manner in the upper region than those in the lower region.

Long-term exposure to high muscle stiffness can lead to muscle fatigue, which in turn increases the risk of developing LBP. Therefore, it is necessary to eliminate asymmetric static and dynamic risk factors, such as leg length discrepancy or abnormal prosthetic gait that can cause LBP in LLAs.

Additionally, it is necessary to provide bilateral symmetrical leg lengths and to maintain the flexibility of the upper back muscles through stretching and strengthening exercises. It is also believed that a continuous monitoring of the back muscles with a simple device such as a hand-held myotonometer will considerably contribute toward preventing LBP in LLAs.

Author Contributions: Conceptualization, methodology, visualization, writing—reviewing, editing, and investigation, Y.C.; conceptualization, methodology, formal analysis, data curation, and investigation, J.K.; supervision, investigation, G.K.; project administration, funding acquisition, H.S. and S.P. All authors have read and agreed to the published version of the manuscript.

Funding: This research was supported by the Bio and Medical Technology Development Program of the National Research Foundation (NRF) funded by the Ministry of Science and ICT (No.2017M3A9E2063255).

Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki, and approved by the Institutional Review Board of Rehabilitation Engineering Research Institute (RERI-IRB-190114-1).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The datasets generated during and/or analysed during the current study are available from the corresponding author on reasonable request.

Acknowledgments: We would like to thank Editage (www.editage.co.kr, accessed on 26 May 2021) for editing and reviewing this manuscript for English language.

Conflicts of Interest: The authors declare no conflict of interest.

References

1. Ehde, D.M.; Smith, D.G.; Czerniecki, J.M.; Campbell, K.M.; Malchow, D.M.; Robinson, L.R. Back pain as a secondary disability in persons with lower limb amputations. *Arch. Phys. Med. Rehabil.* **2001**, *82*, 731–734. [[CrossRef](#)]
2. Ephraim, P.L.; Wegener, S.T.; MacKenzie, E.J.; Dillingham, T.R.; Pezzin, L.E. Phantom pain, residual limb pain, and back pain in amputees: Results of a national survey. *Arch. Phys. Med. Rehabil.* **2005**, *86*, 1910–1919. [[CrossRef](#)]
3. Kušljugić, A.; Kapidžić-Duraković, S.; Kudumović, Z.; Čičkušić, A. Chronic low back pain in individuals with lower-limb amputation. *Bosn. J. Basic Med. Sci.* **2006**, *6*, 67. [[CrossRef](#)]
4. Friberg, O. Biomechanical significance of the correct length of lower limb prostheses: A clinical and radiological study. *Prosthet. Orthot. Int.* **1984**, *8*, 124–129. [[CrossRef](#)]
5. Morgenroth, D.C.; Orendurff, M.S.; Shakir, A.; Segal, A.; Shofer, J.; Czerniecki, J.M. The relationship between lumbar spine kinematics during gait and low-back pain in transfemoral amputees. *Am. J. Phys. Med. Rehabil.* **2010**, *89*, 635–643. [[CrossRef](#)]
6. Devan, H.; Hendrick, P.; Ribeiro, D.C.; Hale, L.A.; Carman, A. Asymmetrical movements of the lumbopelvic region: Is this a potential mechanism for low back pain in people with lower limb amputation? *Med Hypotheses* **2014**, *82*, 77–85. [[CrossRef](#)] [[PubMed](#)]
7. Morgenroth, D.C.; Shakir, A.; Orendurff, M.S.; Czerniecki, J.M. Low-back pain in transfemoral amputees: Is there a correlation with static or dynamic leg-length discrepancy? *Am. J. Phys. Med. Rehabil.* **2009**, *88*, 108–113. [[CrossRef](#)] [[PubMed](#)]
8. Lee, R.Y.; Turner-Smith, A. The influence of the length of lower-limb prosthesis on spinal kinematics. *Arch. Phys. Med. Rehabil.* **2003**, *84*, 1357–1362. [[CrossRef](#)]
9. Schmid, A.B.; Dyer, L.; Böni, T.; Held, U.; Brunner, F. Paraspinal muscle activity during symmetrical and asymmetrical weight training in idiopathic scoliosis. *J. Sport Rehabil.* **2010**, *19*, 315–327. [[CrossRef](#)]
10. Gailey, R.; Allen, K.; Castles, J.; Kucharik, J.; Roeder, M. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. *J. Rehabil. Res. Dev.* **2008**, *45*. [[CrossRef](#)]
11. Burke, M.; Roman, V.; Wright, V. Bone and joint changes in lower limb amputees. *Ann. Rheum. Dis.* **1978**, *37*, 252–254. [[CrossRef](#)] [[PubMed](#)]
12. Butowicz, C.M.; Acasio, J.C.; Dearth, C.L.; Hendershot, B.D. Trunk muscle activation patterns during walking among persons with lower limb loss: Influences of walking speed. *J. Electromyogr. Kinesiol.* **2018**, *40*, 48–55. [[CrossRef](#)]

13. Shojaei, I.; Hendershot, B.D.; Wolf, E.J.; Bazrgari, B. Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals. *Clin. Biomech.* **2016**, *32*, 157–163. [[CrossRef](#)] [[PubMed](#)]
14. Lamoth, C.J.; Meijer, O.G.; Daffertshofer, A.; Wuisman, P.I.; Beek, P.J. Effects of chronic low back pain on trunk coordination and back muscle activity during walking: Changes in motor control. *Eur. Spine J.* **2006**, *15*, 23–40. [[CrossRef](#)]
15. Vogt, L.; Pfeifer, K.; Banzer, W. Neuromuscular control of walking with chronic low-back pain. *Man. Ther.* **2003**, *8*, 21–28. [[CrossRef](#)]
16. Winter, D.; Yack, H. EMG profiles during normal human walking: Stride-to-stride and inter-subject variability. *Electroencephalogr. Clin. Neurophysiol.* **1987**, *67*, 402–411. [[CrossRef](#)]
17. van der Hulst, M.; Vollenbroek-Hutten, M.M.; Rietman, J.S.; Schaake, L.; Groothuis-Oudshoorn, K.G.; Hermens, H.J. Back muscle activation patterns in chronic low back pain during walking: A “guarding” hypothesis. *Clin. J. Pain* **2010**, *26*, 30–37. [[CrossRef](#)]
18. Nair, K.; Masi, A.T.; Andonian, B.J.; Barry, A.J.; Coates, B.A.; Dougherty, J.; Schaefer, E.; Henderson, J.; Kelly, J. Stiffness of resting lumbar myofascia in healthy young subjects quantified using a handheld myotonometer and concurrently with surface electromyography monitoring. *J. Bodyw. Mov. Ther.* **2016**, *20*, 388–396. [[CrossRef](#)] [[PubMed](#)]
19. Campbell, K.; Lakie, M. A cross-bridge mechanism can explain the thixotropic short-range elastic component of relaxed frog skeletal muscle. *J. Physiol.* **1998**, *510*, 941–962. [[CrossRef](#)]
20. Lakie, M.; Robson, L.G. Thixotropic Changes in Human Muscle Stiffness and The Effects of Fatigue. *Q. J. Exp. Physiol.* **1988**, *73*, 487–500. [[CrossRef](#)]
21. Koppenhaver, S.; Gaffney, E.; Oates, A.; Eberle, L.; Young, B.; Hebert, J.; Proulx, L.; Shinohara, M. Lumbar muscle stiffness is different in individuals with low back pain than asymptomatic controls and is associated with pain and disability, but not common physical examination findings. *Musculoskelet. Sci. Pract.* **2020**, *45*, 102078. [[CrossRef](#)]
22. Benjamin, M.; Toumi, H.; Ralphs, J.; Bydder, G.; Best, T.; Milz, S. Where tendons and ligaments meet bone: Attachment sites (‘entheses’) in relation to exercise and/or mechanical load. *J. Anat.* **2006**, *208*, 471–490. [[CrossRef](#)]
23. Andonian, B.J.; Masi, A.T.; Aldag, J.C.; Barry, A.J.; Coates, B.A.; Emrich, K.; Henderson, J.; Kelly, J.; Nair, K. Greater resting lumbar extensor myofascial stiffness in younger ankylosing spondylitis patients than age-comparable healthy volunteers quantified by myotonometry. *Arch. Phys. Med. Rehabil.* **2015**, *96*, 2041–2047. [[CrossRef](#)] [[PubMed](#)]
24. Hu, X.; Lei, D.; Li, L.; Leng, Y.; Yu, Q.; Wei, X.; Lo, W.L.A. Quantifying paraspinal muscle tone and stiffness in young adults with chronic low back pain: A reliability study. *Sci. Rep.* **2018**, *8*, 1–10. [[CrossRef](#)]
25. White, A.; Abbott, H.; Masi, A.T.; Henderson, J.; Nair, K. Biomechanical properties of low back myofascial tissue in younger adult ankylosing spondylitis patients and matched healthy control subjects. *Clin. Biomech.* **2018**, *57*, 67–73. [[CrossRef](#)]
26. Marusiak, J.; Jaskólska, A.; Koszewicz, M.; Budrewicz, S.; Jaskólski, A. Myometry revealed medication-induced decrease in resting skeletal muscle stiffness in Parkinson’s disease patients. *Clin. Biomech.* **2012**, *27*, 632–635. [[CrossRef](#)]
27. Chuang, L.-I.; Wu, C.-y.; Lin, K.-c.; Lur, S.-y. Quantitative mechanical properties of the relaxed biceps and triceps brachii muscles in patients with subacute stroke: A reliability study of the myoton-3 myometer. *Stroke Res. Treat.* **2012**, *2012*, 617694. [[CrossRef](#)] [[PubMed](#)]
28. Bizzini, M.; Mannion, A.F. Reliability of a new, hand-held device for assessing skeletal muscle stiffness. *Clin. Biomech.* **2003**, *18*, 459–461. [[CrossRef](#)]
29. Nourbakhsh, M.R.; Arab, A.M. Relationship between mechanical factors and incidence of low back pain. *J. Orthop. Sports Phys. Ther.* **2002**, *32*, 447–460. [[CrossRef](#)] [[PubMed](#)]
30. Merskey, H.E. Classification of chronic pain: Descriptions of chronic pain syndromes and definitions of pain terms. *Pain* **1986**, *3*, 226.
31. Sakai, F.; Ebihara, S.; Akiyama, M.; Horikawa, M. Pericranial muscle hardness in tension-type headache: A non-invasive measurement method and its clinical application. *Brain* **1995**, *118*, 523–531. [[CrossRef](#)]
32. Myoton. *Hand Holding Myoton Pro User Manual*; Myoton Ltd.: Tallinn, Estonia, 2013.
33. Bruno, A.G.; Burkhardt, K.; Allaire, B.; Anderson, D.E.; Bouxsein, M.L. Spinal loading patterns from biomechanical modeling explain the high incidence of vertebral fractures in the thoracolumbar region. *J. Bone Miner. Res.* **2017**, *32*, 1282–1290. [[CrossRef](#)] [[PubMed](#)]
34. Lee, H.S.; Shim, J.S.; Lee, S.T.; Kim, M.; Ryu, J.S. Facilitating effects of fast and slope walking on paraspinal muscles. *Ann. Rehabil. Med.* **2014**, *38*, 514. [[CrossRef](#)] [[PubMed](#)]
35. Lo, W.L.A.; Yu, Q.; Mao, Y.; Li, W.; Hu, C.; Li, L. Lumbar muscles biomechanical characteristics in young people with chronic spinal pain. *BMC Musculoskelet. Disord.* **2019**, *20*, 1–9. [[CrossRef](#)] [[PubMed](#)]
36. Jubany, J.; Danneels, L.; Angulo-Barroso, R. The influence of fatigue and chronic low back pain on muscle recruitment patterns following an unexpected external perturbation. *BMC Musculoskelet. Disord.* **2017**, *18*, 161. [[CrossRef](#)]
37. Alsancak, S.; Sener, G.; Erdemli, B.; Ogun, T. Three dimensional measurements of pelvic tilt in trans-tibial amputations: The effects of pelvic tilt on trunk muscles strength and characteristics of gait. *Prosthet. Orthot. Int.* **1998**, *22*, 17–24. [[CrossRef](#)]
38. Anderson, F.C.; Pandy, M.G. Individual muscle contributions to support in normal walking. *Gait Posture* **2003**, *17*, 159–169. [[CrossRef](#)]
39. Ceccato, J.-C.; De Sèze, M.; Azevedo, C.; Cazalets, J.-R. Comparison of trunk activity during gait initiation and walking in humans. *PLoS ONE* **2009**, *4*, e8193. [[CrossRef](#)]

40. Kim, D.H.; Jung, Y.J.; Song, Y.E.; Hwang, D.H.; Ko, C.-Y.; Kim, H.S. 1C5-1 Measurement of Characteristic Change using MyotonPRO in Low Back Muscles during a Long-term Driving; Pilot Study. *Jpn. J. Ergon.* **2015**, *51*, S450–S453. [[CrossRef](#)]
41. Kim, C.; Kim, M. Mechanical properties and physical fitness of trunk muscles using Myoton. *Korean J. Phys. Educ.* **2016**, *55*, 633–642.
42. Crawford, R.; Gizzi, L.; Dieterich, A.; Ni Mhuiris, Á.; Falla, D. Age-related changes in trunk muscle activity and spinal and lower limb kinematics during gait. *PLoS ONE* **2018**, *13*, e0206514. [[CrossRef](#)] [[PubMed](#)]
43. Smith, H.E.; Anderson, D.G.; Vaccaro, A.R.; Albert, T.J.; Hilibrand, A.S.; Harrop, J.S.; Ratliff, J.K. Anatomy, biomechanics, and classification of thoracolumbar injuries. In *Seminars in Spine Surgery*; WB Saunders: Philadelphia, PA, USA, 2010; Volume 22, pp. 2–7. [[CrossRef](#)]
44. Hagg, G. Static work loads and occupational myalgia—a new explanation model. *Electromyogr. Kinesiol.* **1991**, *949*, 141–144.