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Changes in Human Leg Function in vertical jumping with and without compliant foot prosthesis

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Solemn Declaration

I hereby declare that this thesis is written by me. Each part, literally or by content, which were published by other sources or authors were marked as references. I only used the sources which were mentioned at the end of this thesis.

Darmstadt, den 3. November 2014 Dennis Dahms

Abstract

Simplified movements like hopping in place are common to extract basic mechanisms in neuromuscular coordination using spring-mass dynamics. In response to a mechanical change in human movement, the musculoskeletal, musculuoarticular and multijoined system has to adapt to the new environment. The goal of this study was to examine the change in human leg function during hopping in place in vertical jumping. One participant performed two-legged hopping with and without spring-legs. During each trial a perturbation platform moves either up (+2 cm) or down (-2 cm, -4 cm, -6cm) to an unexpected moment. The overall leg stiffness increased for all trials before to after the perturbation. Beyond the peak knee moment increases, while the knee angular displacement decreases, this results in significant higher knee stiffness after the perturbation in relation to before. The knee joint compensates the energy addition generated by moving the platform down during the flight phase. Because of its dominating eccentric muscle properties, the knee joint compensates the perturbation and stabilizes the human system.

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1 Introduction

Human locomotion is essential to move our body through space, not even in everyday life situations and in sports. Hence, many Researchers attempted to get insight into the concepts of human movement during the last decades. Several studies, with healthy and disabled humans, were conducted to investigate the biomechanical effects of economical locomotion (Andersson, Andriacchi & Galante, 1981; Chao, Laughman, Schneider & Stauffer, 1983; Györy, Chao & Stauffer, 1976; Riskowski, 2010). In order to examine specific patterns of human gait it is crucial to pinpoint the basic effects from the multijoined, musculoskeletal and musculoarticular system. This approach led to groundbreaking discoveries, like that the function of the leg during running, hopping or walking behaves like a spring as shown e.g. by Alexander (1990), Blickhan (1989), Farley, Glasheen & McMahon (1993) and McMahon and Cheng (1990).

“The advantage of the spring-mass model is its simplicity in studying the mechanical behavior of the musculoskeletal system by using just one spring” (Kuitunen, Komi & Kyrolainen, 2002, p.166). The body mass of the human is affected as a point mass which is attached to the leg spring (see Fig. 1.1) (Alexander, 1988; Blickhan, 1989; Farley, Blickhan, Saito & Taylor, 1991; Farley et al., 1993; Ferris, Louie & Farley, 1998; He, Kram & McMahon, 1991; McMahon & Cheng, 1990). In fact, the gait of the human is realised in a bouncing manner, so that the spring-mass model accesses the overall musculoskeletal behavior in an isolated way (Ferris & Farley, 1997, p.15).

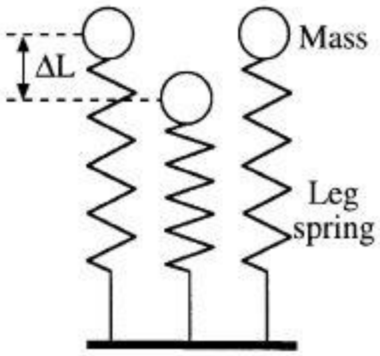


Figure 1.1: Spring-mass Model
The leg is represented as a single spring and the body mass is depicted as a point mass attached to the Leg spring. The ground contact phase is depicted in three different steps (1) start (left-most position) (2) middle (middle position) (3) end (right-most position). The distance ΔL simulates the compression of the Leg spring (Farley et al., 1999, p.268).

This knowledge can be utilised to calculate the stiffness of the leg or joint to observe the foundation of our movement. The "stiffness is the relationship between the deformation of a body and a given force" (Butler, Crowell & Davis, 2003, p.511). As you can see in Fig. 1.1 the deformation is depicted as ΔL which is due to compression of the leg. Various studies occupied this model to investigate the human system and to gain more information in control mechanisms regulated by the central nervous system. Thus, the stiffness is not only the essential element in medical studies (Butler et al., 2003; Granata, Padua & Wilson, 2001; Radin, Ehrlich, Chernack, Abernethy, Paul & Rose, 1978), but also in performance related studies (Alexander, 1990; Kuitunen et al., 2002; Seyfarth, Friedrichs, Wank & Blickhan, 1999). Farley et al. (1993) and He et al. (1991) observed that during forward speed the stiffness of the leg spring remains fairly constant. Moreover, many researches emphasized the dependency between the stiffness and the activity. The result is a higher load of stiffness in order to adapt to the demand of the activity (Arampatzis, Brüggemann & Metzler, 1999; Arampatzis, Brüggemann & Klapsing, 2001a; Arampatzis, Schade, Walsh & Brüggemann, 2001b; Farley et al., 1991; Granata et al., 2001; Kuitunen et al., 2002; Seyfarth, Geyer, Günther & Blickhan, 2002; Stefanyshyn & Nigg, 1998). Another effector of the stiffness is the speed, because a higher velocity leads to a stiffer leg (Arampatzis et al., 1999; Seyfarth et al., 2002). Furthermore the running economy is going to be better with an increase in leg stiffness (McMahon & Cheng, 1990; Kerdok, Biewener, McMahon, Weyand & Herr, 2002).

Recent studies also concluded the relationship between stiffness and performance during hopping in place (Farley et al., 1991; Farley, Houdijk, Van Strien & Louie, 1998; Farley & Morgenroth, 1999; Ferris & Farley, 1997; Haeufle, Grimmer & Seyfarth, 2010; Kalveram, Haeufle, Grimmer & Seyfarth, 2010). The advantage of observing our complex human musculoskeletal system during hopping in place is "… because it follows the same basic mechanics and spring mass model as forward running (Farley et al., 1991) yet has simpler kinematics" (Ferris & Farley, 1997, p.16). Outcomes from this approach were a higher amount of leg stiffness with increased hopping height (Farley et al., 1991; Farley & Morgenroth, 1999) and also with increased hopping frequency (Farley et al., 1991). The major regulator for the leg stiffness during submaximal hopping is the ankle stiffness (Farley & Morgenroth, 1999), but during maximal hopping the knee stiffness is the major determinant (Hobara, Muraoka, Omuro, Gomi, Sakamoto, Inoue & Kanosue, 2009). These studies make clear which joints take control of our movement, though it is still unproven which joints compensate these functions after a mechanical diversification, for example after an amputation. Studies focussing on these aspects were conducted for different locomotions like running (Bruggemann, Arampatzis, Emrich & Potthast, 2008; Hobara, Baum, Kwon, Miller, Ogata, Kim & Shim, 2013; Lechler & Lilja, 2008; Weyand, Bundle, McGowan, Grabowski, Brown, Kram & Herr, 2009) and long jumping (Nolan & Lees, 2007; Nolan, Patritti & Simpson, 2012). The result from Paralympics long jumper Markus Rehm during the German Championship pushes the issue more in the foreground. He won the tournament as amputee against non-amputees, hence the crux is to know, whether long jumpers benefit from their spring-legs or not (Stern.de, 2014).

Observing the human locomotion during irregular perturbation on the ground is a good way to extract the properties of stabilizing the musculoskeletal system (Haeufle et al., 2010; Müller, Grimmer & Blickhan, 2010). Therefore this study focused on the adjustments of the joints during hopping with spring-legs under the circumstances that the ground perturbs up or down and the ankle joint is blocked in relation to hopping without spring-legs. The following hypotheses were examined:

H1: During two-legged hopping in place with spring-legs, the stiffness of the knee is increased when the perturbation platform drops down.

H2: The stiffness of the knee is increased in both cases of two-legged hopping with and without spring-legs after the perturbation platform drops down.

2 Methods

2.1 Participants

Two healthy sport students [24 ± 1 years of age, height 1.71 ± 0.01 m, weight 71 kg] participated in the study. The participants were selected from a pool of potential participants based on his height meeting the experimental constraints. Due to limited space above the head of the participants, a maximum height of 1.75 m was sought. The participants were informed of the potential risks of the study and provided informed written consent prior to participation in the study. The research was conducted according to the declaration of Helsinki and was approved by the ethical committee of the University of TU Darmstadt. One participant withdrew from the study due to schedule conflicts.

2.2 Procedure

Data were collected during one, five hour session. He had no experience in walking or jumping with those spring-legs (see Figure 2.1), so an individual warm-up and training of fifteen to twenty minutes were realised. After this period, the participant had enough self-confidents to perform hopping with the associated spring-legs. In order to get a relation to the specific individual force generation, a test battery for maximum voluntary contraction of different muscles (Gastrocnemius, Rectus Femoris, Biceps Femoris, Soleus, Gluteus Maximus) were determined.

Figure 2.1: Spring-Leg GT Cobra - T-Rex

The subject performed at first two-legged hopping with spring-legs and then without spring-legs. During hopping the platform was moved up (2 cm) or down (2cm, 4cm or 6cm) randomly chosen to an unexpected moment. In order to ensure stable posture, two bars were fixed in front of the participant so he could stabilize himself if necessary. Finally to take precaution the subject donned a climbing harness. It was connected via a rope to a steel rope above his head. In case of a drop it prevents the subject hitting the ground. A mark on the platform was the starting point for every passage. The participant chose his own preferred hopping frequency. A trial was accepted if the platform moved up or down during the flight phase and each height was finished after five proper retries. The participant was not informed which trials were accepted in order to keep the participant naive to the manipulation of interest.

2.3 Material

Kinematics and dynamics data were collected using a Motion Tracking System (Qualisys Software and Hardware, Sampling rate of 500 Hz, 10 cameras, QTM Version: 2.8). The reflective markers were fixed at different body landmarks (left forehead, right forehead, cervical seven, left shoulder, right shoulder, sacrum, left anterior spine of ischium, right anterior spine of ischium, left great trochanter, right great trochanter, left knee lateral, right knee lateral, left ankle lateral, right ankle lateral, left top of the blade, left middle part of the blade, left bottom of the blade, right top of the blade, right middle part of the blade, right bottom of the blade).
Additionally the ground reaction force was collected using a force platform (Kistler force plate) which is movable upwards and downwards with a sampling rate of 1000 Hz. Further on the muscle activity was measured electromyography (Delsys) for multiple muscles (quadriceps femoris, semitendinosus, vastus medialis, vastus lateralis, biceps femoris, gluteus maximus, gastrocnemius medialis, soleus) bilaterally at a sampling rate of 2000 Hz. The recoil of the spring-legs is optimized for a mass between 70^{th} and 90^{th} kg and its own weight is 4.05 kg. Electromyography and motion capturing were triggered using a light barrier and the force signal was started manually by hand. The experimental setup is depicted in Figure 2.2 with all details and components as described above.

Figure 2.2: Setup of the measurement without spring-legs

2.4 Data Processing

In order to generate clear results the raw data was processed using several different algorithms in MATLAB & Simulink (Version: R2012a). Hence to get evidently muscle potentials, peaks had to be extracted at first for each muscle. Subsequently applying a high-pass, low-pass and notch-Filter leads to significant wave lines, which were rectified afterwards. Finally a moving-average-Filter generates the progression of the innervation of the muscle.

A drift in Motion Capturing data and Force data had to be compensated by mapping both together. The force data had to be shifted by a calculated offset, which was measured observing the first contact on the force plate with the given kinematics.

Computing the stiffness of a joint (k_{joint}) was done by dividing the net muscle moment (ΔM_{joint}) with the joint angular displacement ($\Delta\sigma_{joint}$). Net muscle moments were calculated using an inverse dynamic technique which is depicted in Figure 2.3 and described by Chowdhury & Kumar (2013). For hopping with spring-legs the center of mass from the spring-leg were estimated and for hopping without spring-legs the center of pressure was chosen to create a segment with the ankle.

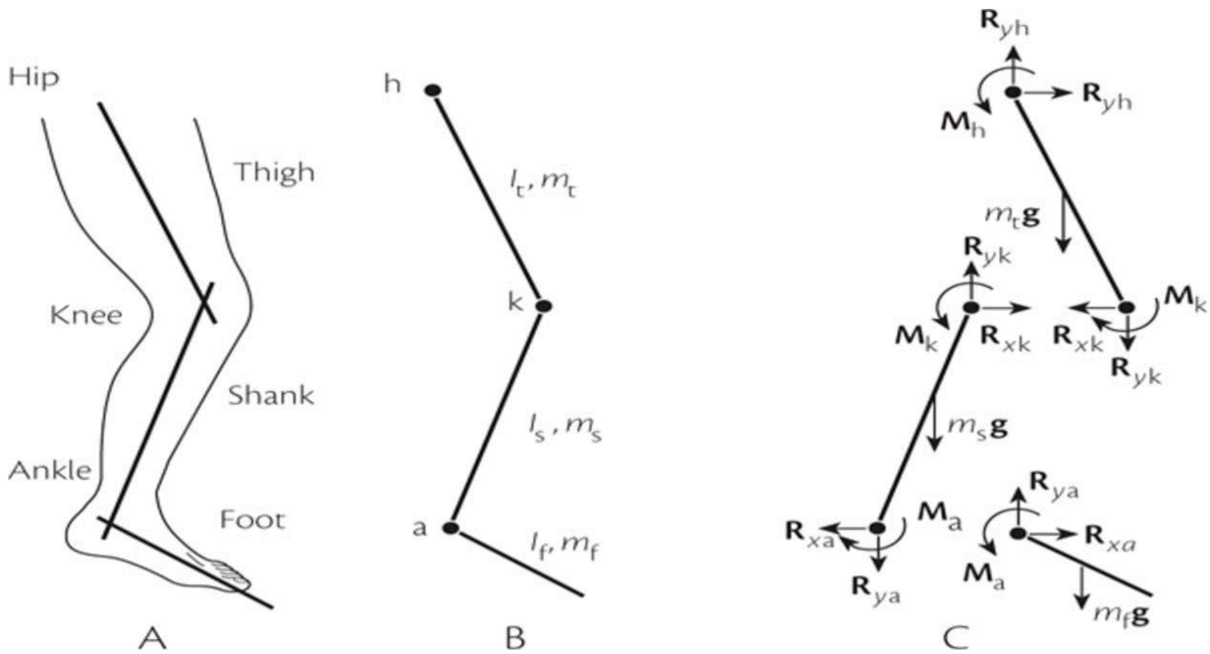


Figure 2.3: Inverse Dynamic approach to calculate the torque of each leg joint

(A) Anatomical model; (B) Link segment model; (C) Free body diagram of lower limb (Chowdhury & Kumar, 2013, p.93).

Position and force data were processed applying a second-order Butterworth filter with a cut-off frequency of 10 Hz and a sampling rate of 500 Hz, so the force data had to be downsampled from 15000 frames to 7500 frames.

Data were rehashed only for the instance when the perturbation platform drops due to a disorder of the perturbation platform caused by an unintended movement of the plate. Consequently the 8th hop before the perturbation was taken into account, because stable hopping was present. Two hops immediately after the perturbation were also taken into account and averaged for all trials with the associated height.

3 Results

3.1 Consideration of force data and marker position

Figure 3.1 and Figure 3.2 displaying the progression of the force data and the cervical 7 marker in case of hopping with and without spring-legs with respect to the time in frames (500 Hz). The negative value in the force data indicates the movement of the force plate associated with the given heights. In each trial the amount of force increases significantly after the perturbation and decreases with further hops. The maximum peak force divers for each trial dependent of the height of perturbation. The hopping height (simulated by the red line) is approximately stable unless right after the perturbation, where a higher hopping height is present. In case of hopping without spring-legs (Fig. 3.2) an increase in peak force is also prevalent and is much higher than in hopping with spring-legs. Normally the marker position should change immediately after the perturbation (Fig. 3.2A, Fig. 3.2B) and holds the position but a downward movement before the perturbation is present (Fig. 3.2C). Further the marker position moves also down for the second hop after the perturbation (Fig. 3.2A, Fig. 3.2B).[1]

[1] Due to a high force impact the motor could not stabilize the position of the force plate. Hence a downward movement was the reason why the marker position changed.

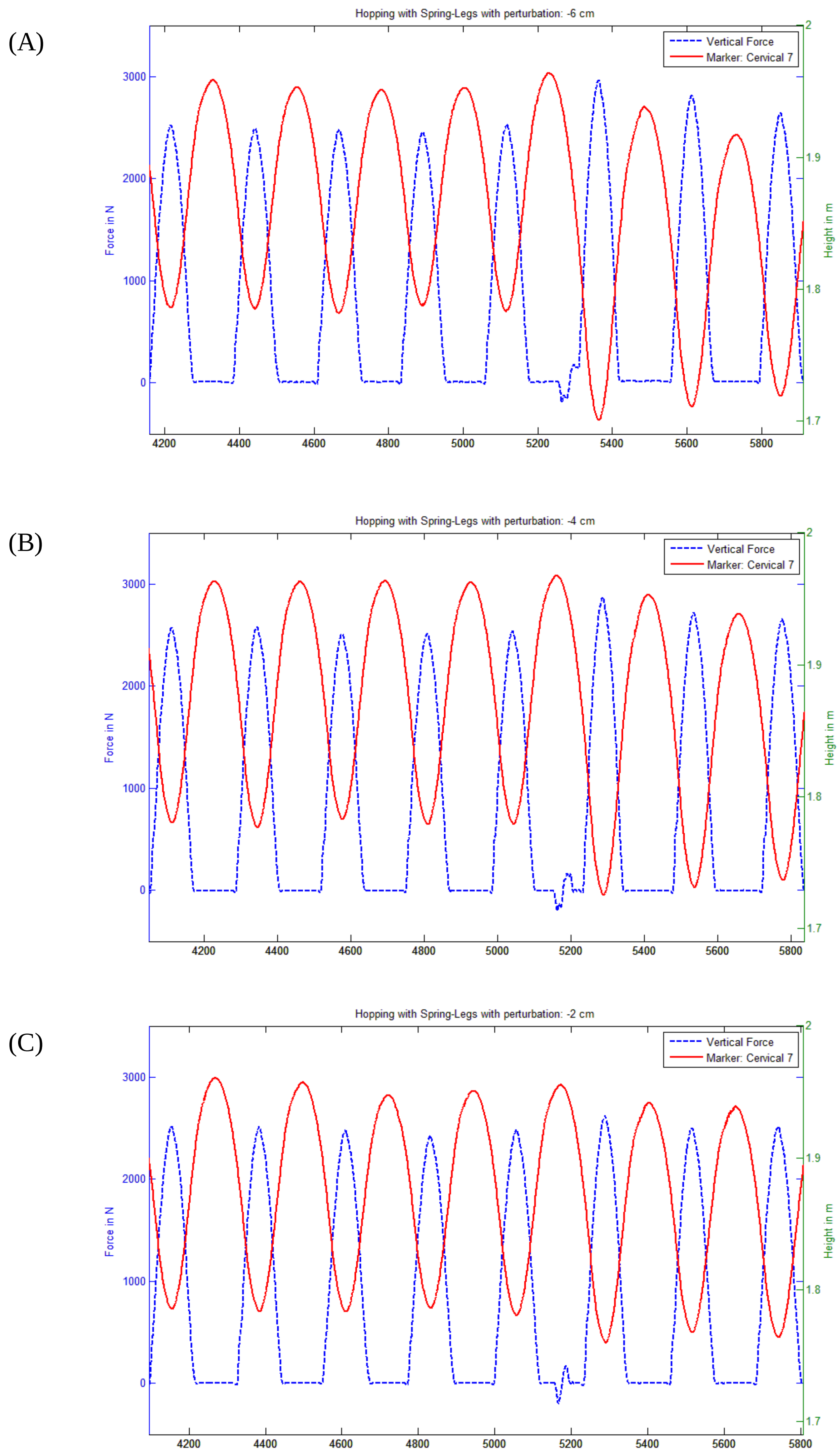
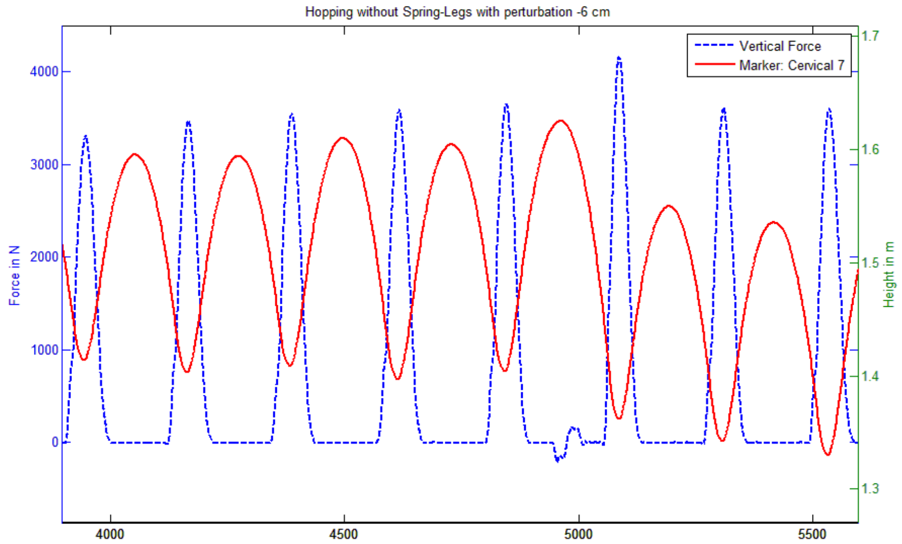


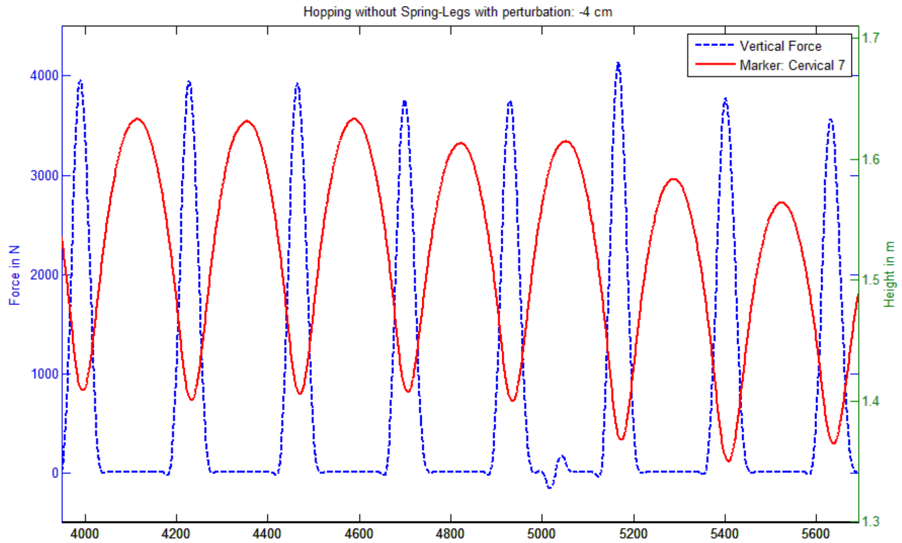
Figure 3.1: Hopping with spring-legs for different perturbations

(A) Perturbation: -6 cm (B) Perturbation: -4 cm (C) Perturbation: -2 cm

(A)



(B)



(C)

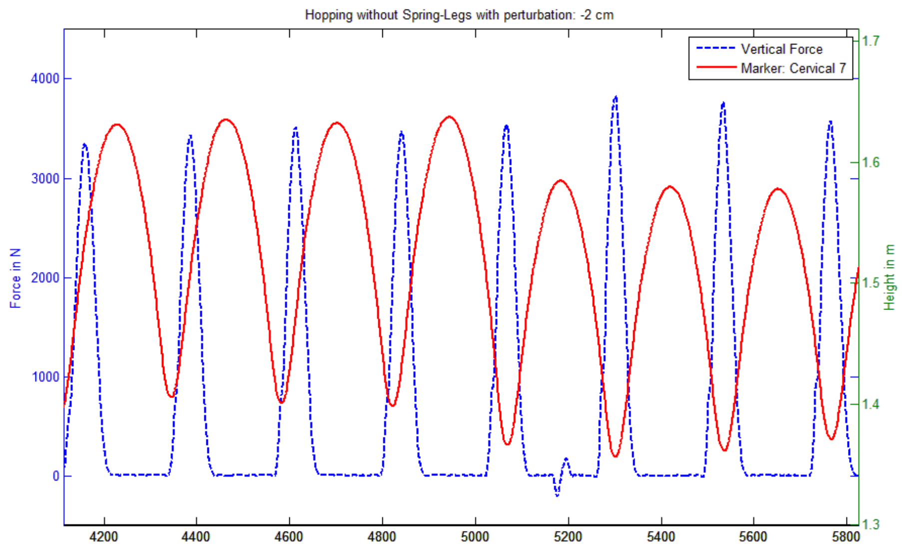


Figure 3.2: Hopping without spring-legs for different perturbations

(A) Perturbation: -6 cm (B) Perturbation: -4 cm (C) Perturbation: -2 cm

3.2 Hopping parameters

Table 3.1: Overwiew of parameters for hopping with spring-legs for different perturbation heights. Chosen data is separated before and after the perturbation. Standard deviations are presented in parenthesis.

	Minus six		Minus four		Minus two	
	before	after	before	after	before	after
Peak GRF (N)	2596 (74)	2944 (42,6)	2700 (190)	2884 (196)	2840 (263,1)	2956 (255,4)
Knee peak moment (Nm)	119 (26,3)	150 (22,8)	93 (32,1)	125 (36,2)	89 (15,8)	107 (23,1)
Knee angular displacement (rad)	0,256 (0,05)	0,183 (0,02)	0,240 (0,01)	0,187 (0,04)	0,247 (0,04)	0,200 (0,06)
Knee Stiffness ($Nm\,rad^{-1}$)	555 (258)	811 (132)	390 (136)	671 (118)	371 (116)	556 (118)
Max. Leg Stiffness ($kN\,m^{-1}$)	13,1 (0,8)	15,4 (1,5)	12,3 (1,2)	14,0 (1,9)	12,3 (1)	13,6 (1,6)
Duty-Factor	0,52	0,46	0,52	0,47	0,59	0,54
Ground contact time (s)	0,226 (0,006)		0,220 (0,045)		0,218 (0,004)	
Hopping frequency (Hz)	2,2		2,7		2,5	

The participant increased the vertical force and knee peak moment before and after the perturbation. Additionally the knee angular displacement decreased for each height before and after the perturbation, so the knee stiffness increased by 1.46-fold for the height minus six, 1.72-fold for the height minus four and 1.5-fold for the height minus two. Knee stiffness increased significantly in case of a higher downward movement, this is also prevalent for the leg stiffness. Here, the leg stiffness increased before and after the perturbation and also for a higher amount of downward movement. Other hopping experiments collected data for the knee stiffness with up to 631 $Nm\,rad^{-1}$ and leg stiffness up to 29.3 $kN\,m^{-1}$ (Farley & Morgenroth, 1999, p. 270). Knee angular displacement differs also in relation to other experiments, e.g. during preferred hopping with an angular displacement of 0.396 rad.

The height of perturbation affected also the duty factor whereas a high perturbation leads to a smaller duty factor (0.46) in relation to a lower perturbation (0.54) with a higher duty factor. However, the ground contact times are quite similar regardlessly of the disturbance. The hopping frequency changed for all trials but not in response to a given perturbation height (Table 3.1). While during hopping without spring-legs the individual hopping frequency is stable (Table 3.2). Also the contact time decreases significantly for hopping without spring-legs in comparison to hopping with spring-legs. The duty factor also adapts to the situation before and after the perturbation (Table 3.2) and is lower for all trials in relation to hopping with spring-legs.

Table 3.2: Overwiew of parameters for hopping without spring-legs for different perturbation heights. Chosen data is separated before and after the perturbation[2]. Standard deviations are presented in parenthesis.

	Minus six		Minus four		Minus two	
	before	after	before	after	before	after
Duty Factor	0,48	0,38	0,42	0,38	0,44	0,40
Contact time (s)	0,175 (0,004)		0,171 (0,01)		0,170	
Hopping Frequency (Hz)	2,2		2,2		2,3	

[2] Data processing for hopping without spring-legs was not possible, because of a drift in motion capturing and force data. Hence, with inconsistent data it was not feasible to apply inverse dynamics, which requires reliable data to calculate for example joint moments. Furthermore the electromyography data could not be highlighted because no reference data is present for hopping.

4 Discussion

The goal of this study was to observe the change in human leg function in vertical jumping. In this case the knee function during hopping in place with spring-legs was observed and processed. Mechanical variations are possible on condition of an amputation, hence our human biological system has to adapt to the new situation. Deriving from this requirements two hypotheses were defined:

H1: During two-legged hopping in place with spring-legs, the stiffness of the knee is increased when the perturbation platform drops down.

H2: The stiffness of the knee is increased in both cases of two-legged hopping with and without spring-legs after the perturbation platform drops down.

4.1 Comparison to other studies

Different phenomenons were present in the collected data. Figure 3.1 and Figure 3.2 displays the “terrain following strategy” described by Kalveram et al. (2010). Here, it seems that the distance between the ground profile and the maximum hopping height is constant (Kalveram et al., 2010, p. 1). The contact time between hopping with spring-legs and normal hopping is different. This issue is discussed in media because of the recent results from Paralympics long-jumper Markus Rehm during the national Championship in Germany for non-disabled long-jumpers. The storage of elastic energy is generated by a longer contact time, so the spring deflection increases. Decreasing the ground contact time during two-legged hopping leads to a stiffer leg (Farley et al., 1998, p. 1044). Responsible for decreasing the ground contact time is the higher stiffness in the ankle joint (Kuitunen et al., 2002, p. 171).

4.2 Knee stiffness during hopping with spring-legs

Knee stiffness is calculated by the knee peak moment divided with the knee angular displacement. As Table 3.1 reveals the knee stiffness increases significantly from before the perturbation to after the perturbation and also for higher downward directed movement of the force plate. In case of downward movement of “minus six” the knee stiffness increases 46%, in case of “minus two” stiffness increases 72% and for “minus two” an increase up to 50% is prevalent. The higher amount of stiffness is due to a higher peak moment and a smaller change in knee angle. The participant encounters the ground level with a more extended leg than in normal hopping, so the change in knee angle remains quite small. Based on the downward movement of the force plate, the flight phase increases and a higher impact to the musculoskeletal and musculoarticular system is generated. This is the clue why the knee peak moment increases despite a smaller lever arm. Hence, the knee compensates the higher impact because of high eccentric force properties. The knee joint is the actuator for damping the human system. Not even in hopping (Farley & Morgenroth, 1999) but also in running the ankle and knee stiffness adapting to an altering grounds (Müller et al., 2010). Therefore, the data indicates that the hypothesis H1 is proven.

4.3 Knee stiffness during hopping without spring-legs

Due to a drift in force and motion capturing data, no reliable information could be processed, so the hypothesis H2 remains unclear. Based on previous researches (Farley et al., 1999; Ferris & Farley, 1997; Müller et al., 2010) the tendency supports the hypothesis but has to be examined in future to gain more information in the control of our movement.

4.4 Conclusion

In order to gain more information about the basic mechanisms controlling our movement under disturbing circumstances, more data should be methodically collect. Analyzing a bigger population leads to general and robust information. A particular training with the spring-legs should baseline-separate the learning effects. Also the design of the research and the measurement should be improved to extract the influence of the posture from the participants and create a segment body model which is capable of applying inverse dynamics. Further external influences should be controlled so that the perturbation platform stabilizes the position. A trigger is necessary to combine all methods of measurements to one comprehensive data collection. In that case it is possible to gain reliable insight into the basic mechanisms in controlling our human movement. In addition, electromyography can explain the influence of our central nervous system in response to unexpected disturbance. Preactivation could be method for both instances in moving the force plate up or down. Especially an upward movement of the force plate can generate new information into to concepts of stabilizing our musculoskeletal system. A muscle-tendon-model can describe and analyze the response to a perturbation more in detail.

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