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PREDICTION OF CONTACT FORCE AT MEDIAL KNEE POINT

Nawar Banwan Hassan

ABSTRACT: (KCF) Knee contact force is one of important expressive parameters to evaluate function of prediction knee joint. The study aimed to investigate whether the predicted forced will effect on knee joint and how can reduce the force of knee joint pressure while walking and running. The better understanding of the complex mechanism and its resulting joint kinematics allows a better elucidation of the mechanisms behind body growth, aging progression. Data conquest to patient with data of height 168 cm, weight 66.7 kg, knee side of right and shoes are Rockport's flat bottom sneakers. We observed altered muscle force ranges. The result shown that, The overall quantitative assessment revealed an average square root error ranging from 5.5 to 189.1 N. The present study showed an important variation of kinematical and muscle force behaviors under the voluntary altered gait patterns. The researcher conclusion shown that the simulation result of useful information could allow a better understanding of the joint function design and muscle leading to deeper knowledge of joint and muscle mechanisms related to neural commands.

Keywords: Knee contact force, Knee joint, Gait Cycle.

1- Introduction

Knee contact force (KCF) is one of important expressive parameters to evaluate function of prediction knee joint. The effect of KCF is strongly demonstrated in walking patterns (Buchanan, 2013), knee joint alignment. It also strongly affects directly affecting cartilage pressures and knee stiffness (M. Hunt, 2015). KCF was therefore calculated on a broader and comprehensive scale to assess knee joint disease, walk adjustment, prosthetic design and surgery intervention. Especially for rehabilitation of knee, that the knee joint can off-loading slow cartilage damage and postoperative surgery operation we will replace the detailed or lower wear and lengthen the clinical time of the knee prostheses after surgery (K.B. Shelburne, 2013).

The knee joint loading during walking has been evaluated in various ways. External reaction forces (chambers) and joint knee joint forces estimated as a measure of knee joint loading. (Robertson Gordon, Caldwell Graham , Hamill Joseph, Kamen Gary, Whittlessey Saunders, 2014). Others have applied biomechanical models based on reverse dynamics, which include the contribution of the knee muscle and flexor muscles to assess internal knee joint compression and shear cutting. When fitness trainers and doctors should decide whether the walking pole is to be recommended, the arguments for these recommendations should be evidence-based, hence, mechanical loading documents for the joints of the lower limbs either during normal walking and bipolar walking seems relevant. (C. Richards, J.S. Higginson, 2013).

Studies conducted on cane walking as well as walking have indicated that pregnancy transmitted by reed may reduce pregnancy on the knee joint. If this is true, walking level with increased strength can be expected to result in a significant reduction in knee pressure strength. We expect that if the vertical strength component is large enough to help the lower limbs support the body weight during the position and thus prevent the leg from collapsing, this will result in a decrease in the strength of knee joint pressure. If strength is significantly different in previous studies of walking, this may explain the contradictory results on the effect of walking on knee joint loading. (B.J. Fregly, D.D. D'Lima, C.W. Colwell, 2010).

When the term "load" is used in the following, it indicates the strength of knee joint pressure. We assessed the internal loads of the pressure and shear forces of the knee joint. Muscle strengths played an important role in estimating intracranial loading. (A.L. Kinney, 2013).

When one foot is placed on the ground, which makes it connected in any way to the ground, then the gait cycle begins and ends when the same foot is connected again with the ground (Mooney, Medical Dictionary for the Health Professions and Nursing, 2012). Therefore, we can divide the gait cycle to different stages to determine the standard and abnormal walking and we can determine the prediction of the condition of the knee and the moderate impact on the knee. The gait cycle is the interval between the same exact repetitive events of a walk that usually begins when one foot is connected to the ground [although the perfect walking course can be considered from any foot position while walking] (Mooney, Illustrated Dictionary of Podiatry and Foot Science, 2017). (The eight phases of human gait cycle is shown in appendix A)

Figure (1) the eight phases of human gait cycle (www.streifeneder.com, 2017)

In other sides, the electromyography (EMG) is the surface electromyography is commonly used to define the muscle activation pattern of cutting edge computational to estimate the joint kinematics and muscle force. A recent review study showed kinetic and kinematic changes by using some gait modifications such as increases of step width, cadence, knee flexion, mediolateral trunk lean, or weight transfer to the medial knee as well as reductions of vertical acceleration and initial contact or stride length or speed or using gait aids (e.g., ipsilateral/contralateral cane) (D. Staudenmann, 2010).

Problem Statement

The research problem is the problems of resulting from forced influence and its effect on a patient with a knee joint. The study aimed to determine whether the expected forced effect would affect the knee joint and how it could reduce the strength of knee joint pressure during walking and running

We have previously observed that KCF does not lead to a reduction of pregnancy on the knee joint and it is not clear whether increased strength transmitted by the bipolar can reduce the load on the knee joint. Thus, the purpose of this study is to investigate whether the predicted forced will effect on knee joint and how can reduce the force of knee joint pressure while walking and running.

2- Methodology

Study Design

This research is considered an applied research that plans to method involved of three main actions: (i) data pre-processing with down-sampling and regularizing the data, (ii) input variable selection (IVS) to indicate the most applicable inputs for the rational network, and (iii) FFANN estimate of medial KCF established on the designated inputs within three simplification stages.

2-1 Data and Sampling

All used data are briefly summarized here. The sample conquest was performed in a patient as shown below:

1. Height: 168 cm.
2. Weight: 66.7 kg.
3. Knee side: right.
4. Shoes: Rockport’s flat bottom sneakers
5. Gait pattern before: over ground gait pattern.
6. Gait pattern after: Over ground gait trials
7. Kinematical frequency: 120 Hz.
8. Reaction of ground frequency: 1000 Hz.

2-2 Flow of Action

Based on skin-marker trajectories, reverse kinematics were performed for the calculation of knee joint kinematics. Then, inverse dynamics and constant optimization were introduced to estimate the knee muscular forces according to the normal gait. All these algorithms were performed using OpenSIM 3.1 version on a computer. Inverse dynamic aims at estimating net joint moments from tracked segment kinematics. Constant optimization allows the estimation of the muscle forces to be performed using the equilibrium principle between net joint moments, muscle lever arms, and muscle.

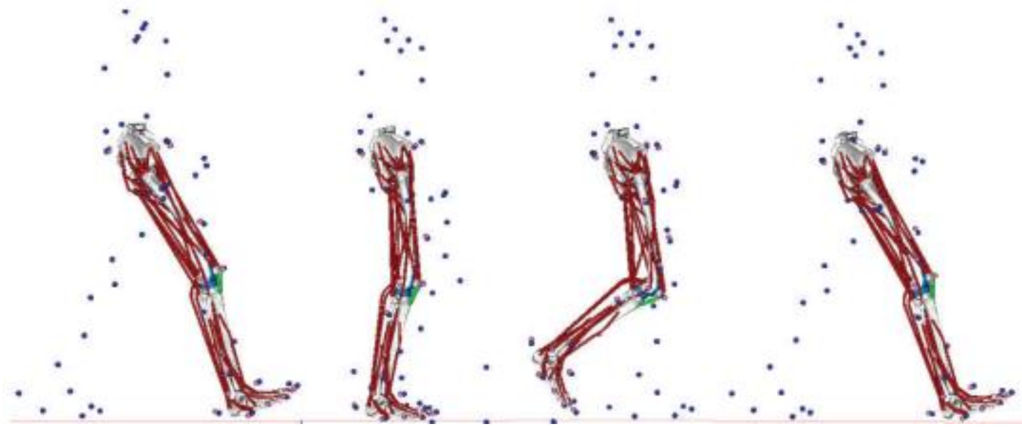

Figure (2) Visualize the movement of a normal gait model patient specific overlapping with experimental skin-56 Portable (blue) and 22 virtual skin Portable (pink color) tag sets . (Lin, 2010)

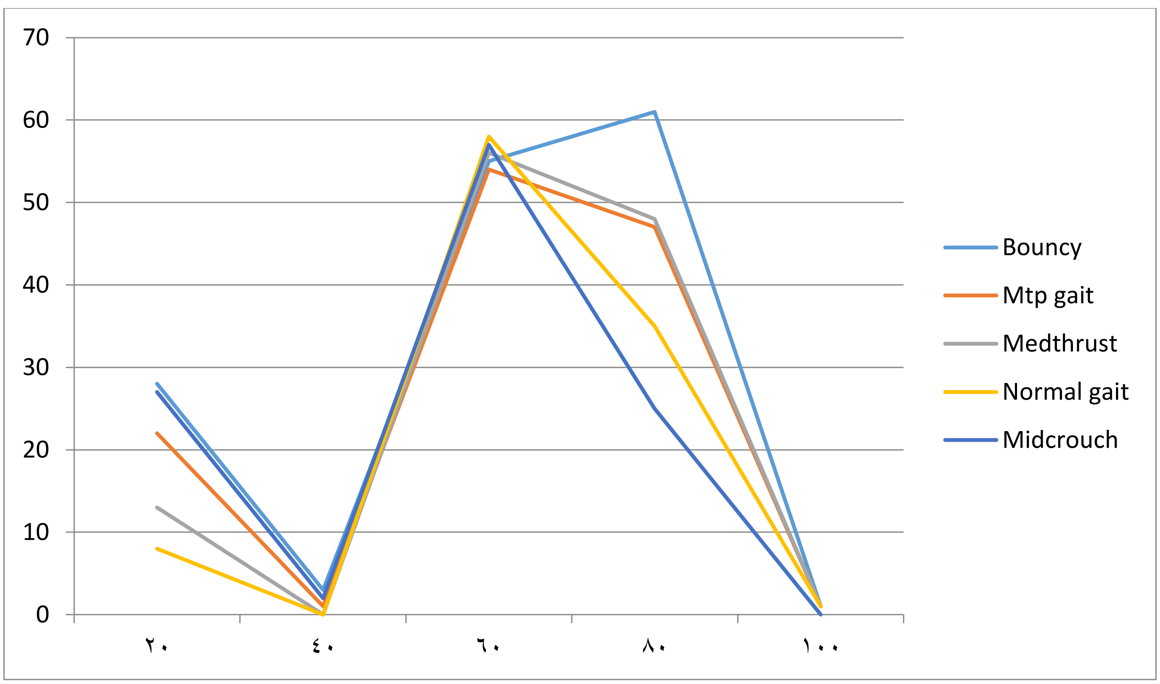

Figure (3) Knee gait cycle (Ana S Machado, 2015)

The next step was to perform a moving average in order to get a clearer signal in the graphs. In the moving average, an interval of 120 was selected. (See appendix C)

To assess the kinematical deviation, a peak-to-peak absolute error was computed for the normal gait pattern. The deviation of muscle forces was quantified using root mean square error (RMSE) and its relative error (RE) according to the maximal value. These quantities were calculated for the normal gait pattern. All post-processing steps were performed using MATLAB R2010b.

Once the data is extracted in SPSS, a component matrix and a was obtained(under results). Analyzing the component matrix below begins with dimension reduction where components with less than 95 % in the 'percent cumulative' column was disregarded as they are not required as principle components (tables2&4,). Component 1 is always the principle and the main component as it has the highest eigenvalue (largest variance). A component with a KMO(Kaiser-Meyer-Olkin) value of less than 0.5 means that a sampling is not adequate. A value close to zero means that there are large partial correlations compared to the sum of correlations. Component 5 in the component matrix (table2, 3, 4 and 5 in appendix B) for example, contains values less than 0.5 and it therefore explains less than one muscle.

The optimization problem is expressed as (I. Moon, 1995)

Minimize F_{abj}

Subject to $R(q)\, F^{TM} = T_{MT}$

Where

- $0 < F^{TM} < F^{0}_{M}$
- q: joint angles set for n joints
- R(q) : muscle moment (n*m) matrix
- F^{TM} : muscle force (m*1) matrix
- T_{MT} : muscular joint moments (n*1) matrix
- F^{0}_{M} : peak isometric muscle force
- F_{abj} : an activation based objective

$$F_{abj} = = \left(\sum_{j=1}^{N} aj\ (ti)\right)2$$

2-3 Ethical considerations

Participants received written and oral information about the purpose of the study, and its way of conducting the test. Participants in the study include the possibility of an additional assessment of knee patients. Ethical approval has been granted by the local Ethics Committee.

3- Data analysis

3-1 Data analysis walking force

Table 1 Data of walking

Method	Data of walking
EMG: Data recording in	2.815 s
kinetic and kinematic	
Total frame	351 frame
Time period	2,823 s

The next stage is a moving average procedure to obtain a clearer signal in the charts. In the moving average a break of 120 was selected.

Muscle force and EMG evolutions of muscle during the normal gait and the medial thrust gaits are shown in Figure 4. Overall, the comparisons revealed a fair qualitative agreement between muscle force and EMG-based patterns of each muscle during the normal gait and medial thrust gait. The same observations were found for the bouncy, midcrouch, and mtp gait modifications. However, at some periods over the gait cycle, the high EMG activity does not perfectly correspond to the muscle peak force.

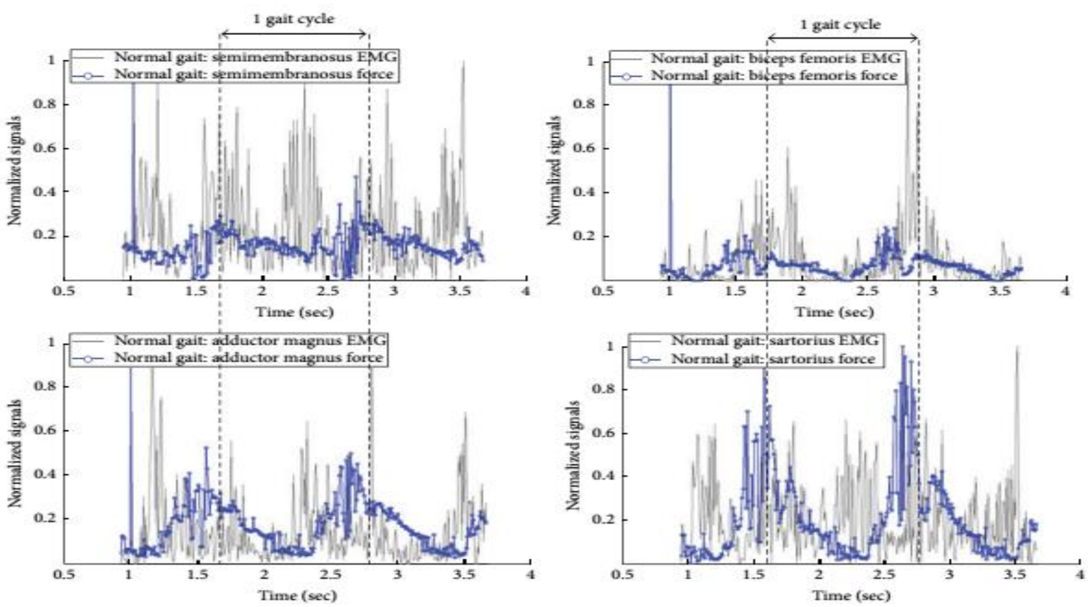

Figure (4) Muscle behaviors during the normal gait and EMG signals

3-2 Gait cycle

While getting ready to move, it is created viable course for a repeat of the movement: This pattern of movement begins and ends with touching the ground of one foot. For example, the right foot and there is also reference foot, start walking cycle when the right foot touches the ground with the heel and ends when it connects to the ground with the toes. It describes the situation phase while the foot is on the ground while the swing stage shows a time when the foot is off the ground. The term 'stride' is known as the movement of each of the parties that occur during the walking cycle.

Root mean square errors range (figure 5) from 7.9 to 139 N, from 7.4 to 189.1 N, from 5.7 to 101.3 N, and from 7.4 to 126.2 N for the comparisons between the normal gait and the bouncy, medial thrust, midcrouch, and mtp gait modifications, respectively. Overall, quantitative assessment revealed a root mean square error ranging from 5.5 to 189.1 N for all comparisons. The maximal deviation arises from the gluteus medius muscle during the medial thrust gait modification. Regarding the relative deviation, the overall range is from 10 to 34%. The medial gastrocnemius revealed a minimal deviation during midcrouch gait modification while the sartorius showed a maximal deviation during the mtp gait modification.

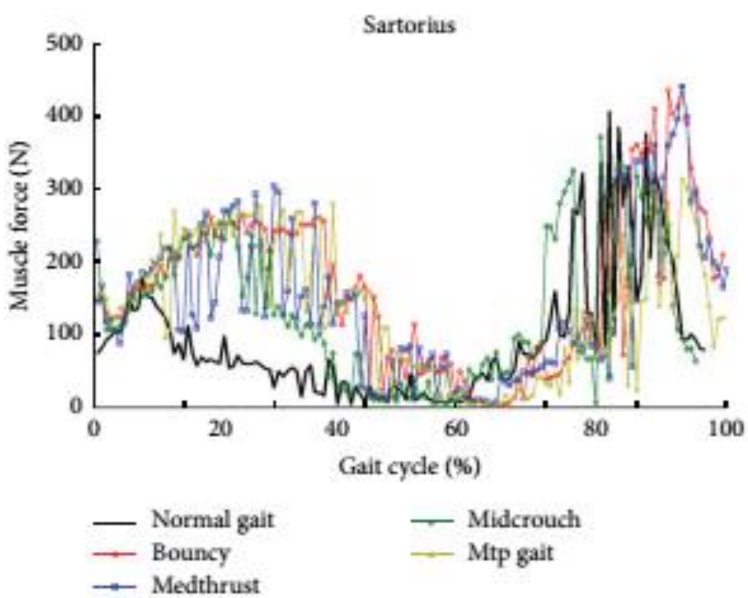

Figure (5) Evolution patterns of the thigh muscle (semimembranosus, biceps femoris, rectus femoris and Sartorius) forces

4- Results

A motion of the patient specific model within the knee implant during his normal gait cycle is shown in figure 2. The duration of his normal gait cycle is around 2.815 sec. The time durations of the bouncy, medial thrust, midcroush, and mtp gaits are around 1.148, 1.156, 1.079, and 1.145 sec. Thus, the development of the knee within his organ is demonstrated during normal and modified brides in Figure 3.

When analyzing the difference between each gait modification according to the normal gait, we observed altered muscle force ranges in figure 5. The overall quantitative assessment revealed an average square root error ranging from 5.5 to 189.1 N for all status. Muscle extracted from the EMG signals during the midcroush gait modifications are illustrated in Figure 6. We noted that the activation levels of the Vastus lateralis and Tibialis anterior are dominant for the midcroush gait.

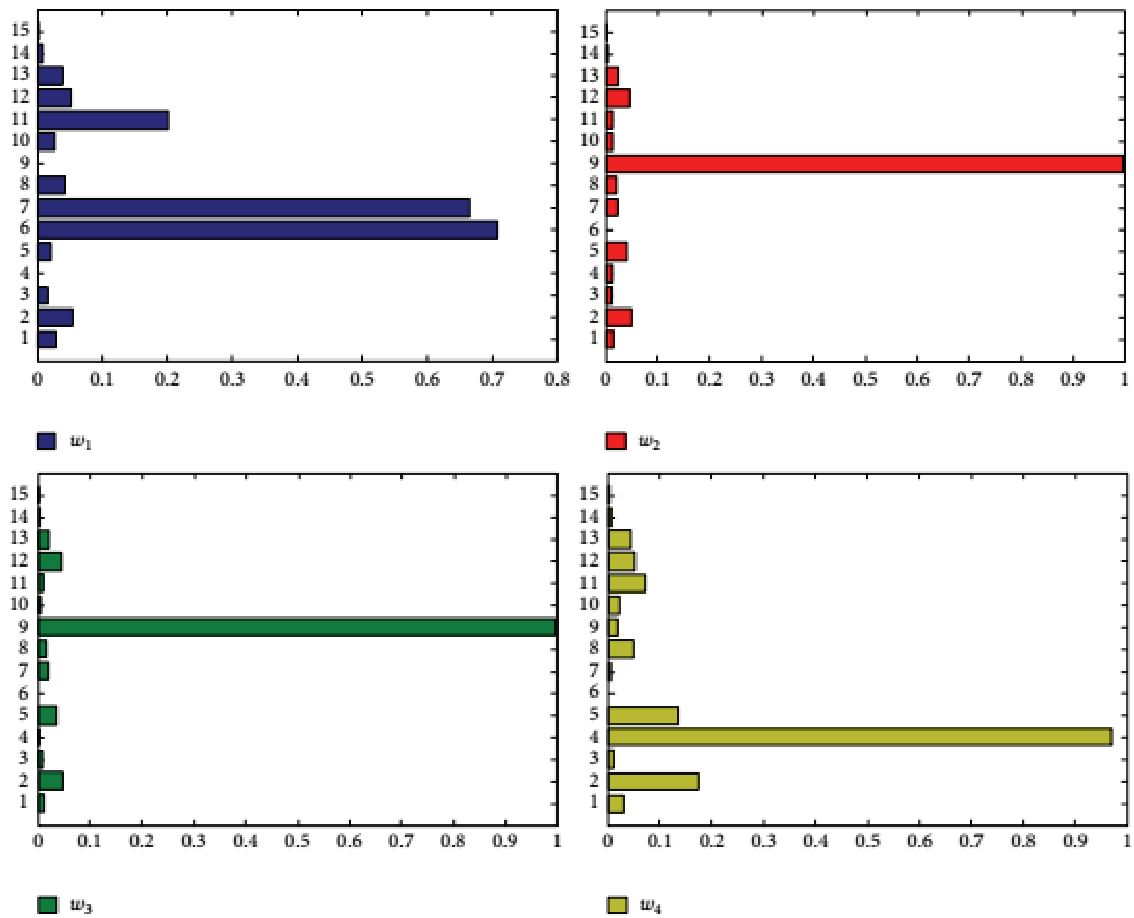

Figure (6) Muscle synergies matrix (w: activation level) indemnified from the nonnegative matrix factorization based on the EMG signals collected during the midcroush gait

Semimembranosus (Choh, 2010)

1- Biceps femoris

2- Vastus medialis

3- Vastus lateralis

4- Rectus femoris
5- Medial gastrocnemius
6- Lateral gastrocnemius
7- Tensor fascia latae
8- Tibialis anterior
9- Peroneus longus
10- Soleus
11- Adductor magrius
12- Gluteus maximus
13- Gluteus medius
14- Sartorius

Each column of muscle synergies (w1, w2, w3, w4) includes a vector of relative level of muscle activations.

5- Discussion

The resulting of joint kinematics and muscle forces provide basic knowledge of locomotion function of the human body in the normal as well as the (EMG) conditions. It is expected that the results obtained from the experiment to provide the advantages or disadvantages of analysis force for knee and prediction of contact force at medial knee point. When walking. However, the moment the knee results in higher electrode data. Therefore, they did not show much benefit. In the results of kinetic data, minor differences can be seen between the regular results and the walking. This tells us that it was used more power on the knee in running walking while normal walking that's used most of the force when we running. Normal walking has helped the infantry to use less power when you go out of the ground with the toes, which is crucial in pushing the body forward.

On the other side, Common kinematics and muscle strength result in the basic knowledge of the human body's motion function in normal conditions, as well as (EMG). Now, cutting-edge computational approaches such as inverse dynamics, static and dynamic optimization, and EMG-driven forward dynamics are commonly used to elucidate the neurophysiological relationships between the motor cortex, the muscle, and the joint mechanics. Each approach has underlying advantages as well as disadvantages. Inverse dynamics and static optimization showed high residual results due to dynamic inconsistencies between Ground Reaction Forces (GFR) and captured kinematics as shown in figure 6. Kinematical errors coming from anatomical representation, marker registration, and soft tissue artifacts may alter the accuracy of the results. Moreover, the inaccuracy may come from the noise in GRF or drift effect over time. Furthermore, the choice of the objective function may create non uniqueness solutions. However, this approach has a short computing time and this may be appropriately suitable for real-time simulation. Forward dynamics and dynamic optimization have computational complexity. The accuracy of EMG signals due to electrode location and signal processing approach may alter significantly the simulation results. Another challenge is how to deal with the recruitment mechanism of units in pathological muscles. Thus, the estimation of muscle forces using optimization deals with no uniqueness cost function and this approach may be invalid for movement disorders. The EMG-driven approach is challenging, but there are many disadvantageous factors related to the cross talk effect and noise or uncertain muscle model parameters.

6- Conclusion

The present study showed an important variation of kinematical and muscle force behaviors under the voluntary altered gait patterns. Our findings allowed also musclesynergies to be recognized for the normal gait as well as 4 gait modifications (bouncy, medial thrust, midcroush and mtp). As a result, useful information can help to better understand knee joint function and muscle synergy that leading to deeper knowledge of the mechanisms of joints and muscles the result.

7- Recommendations

There is a need for future research efforts to participate physics-based analyzes with smart algorithms to perform as a real-time procedures and to estimate joint load operations because of walking adjustments and rehabilitation approaches.

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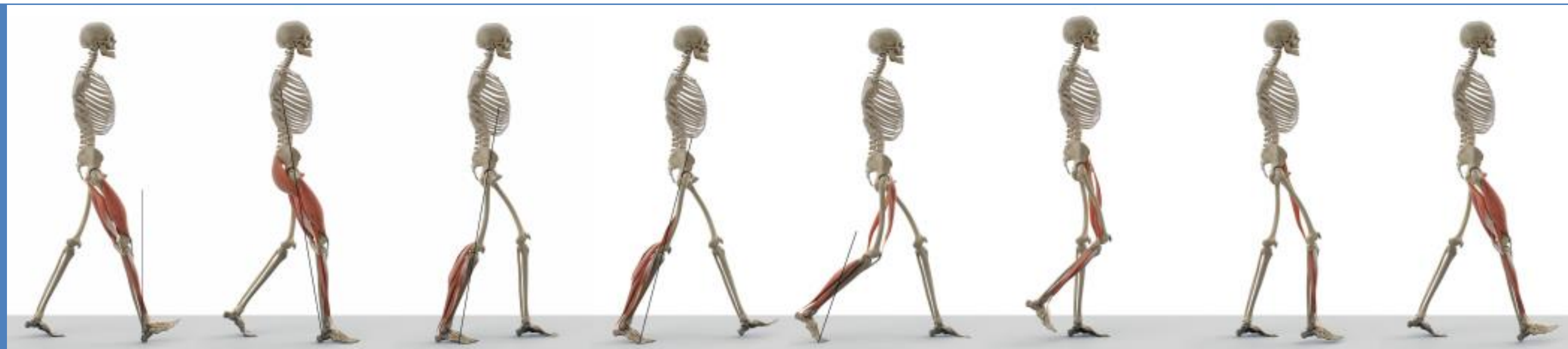
تنبؤ قوة الاتصال في نقطة الركبة للإنسان

الملخص: قوة اتصال الركبة (KCF) هي واحدة من المعاملات التعبيرية الهامة للتقييم والتنبؤ بوظيفة مفصل الركبة. هدفت الدراسة إلى معرفة ما إذا كان التأثير القسري المتوقع سيؤثر على مفصل الركبة وكيف يمكن أن يقلل من قوة ضغط مفصل الركبة أثناء المشي والجري. إن الفهم الأفضل للآلية المعقدة وما ينجم عنها من حركية مشتركة تسمح بتوضيح أفضل للآليات وراء نمو الجسم و التقدم في السن. بيانات المريض (الطول 168سم، الوزن 66.7 كجم)، الركبة اليمني، والأحذية هي أحذية رياضية مسطحة من أسفل Rockport's. لاحظنا تغيراً في نطاقات قوة العضلات. وأظهرت النتائج أن التقييم الكمي الشامل كشف عن متوسط خطأ في الجذر التربيعي يتراوح من 5.5 إلى 189.1 نيوتن. أظهرت الدراسة الحالية تباينًا هامًا في السلوكيات الكينماتية والقوة العضلية تحت أنماط المشية المتغيرة الطوعية. أظهر استنتاج الباحث بأنه يمكن أن تؤدي نتيجة المحاكاة للمعلومات المفيدة إلى فهم أفضل لتصميم الوظيفة المشتركة والعضلات المؤدية إلى معرفة أعمق لآليات المفاصل والعضلات المرتبطة بالأوامر العصبية.

الكلمات المفتاحية: قوة اتصال الركبة، مفصل الركبة، دورة المشي.

Appendix A (www.streifeneder.com, 2017)



Gait phases	IC Initial Contact	LR Loading Response	MST Mid Stance	TST Terminal Stance	PSW Pre Swing	ISW Initial Swing	MSW Mid Swing	TSW Terminal Swing
Gait cycle	0 %	0 – 12 %	12 – 31 %	31 – 50 %	50 – 62 %	62 – 75 %	75 – 87 %	87 – 100 %
Hip	20° flexion	20° flexion	0° flexion	-20° hyperextension	-10° hyperextension	15° flexion	25° flexion	20° flexion
Knee	0° – 5° flexion	20° flexion	0°–5° flexion	0° – 5° flexion	40° flexion	60° – 70° flexion	25° flexion	0° – 5° flexion
Ankle joint	0°	5° – 10° plantar flexion	5° dorsal flexion	10° dorsal flexion	15° plantar flexion	5° plantar flexion	0°	0°
Muscle activity	M. quadrizeps femoris M. tibialis anterior M. gluteus medius M.	M. quadrizeps femoris M. tibialis anterior M. gluteus medius	M. gastrocnemius M. soleus	M. soleus M. gastrocnemius M. flexor digitorum longus M. flexor	M. soleus M. gastrocnemius M. rectus femoris M. adductor longus	M. extensor hallucis longus M. flexor hallucis longus M. sartorius M.	M. semimembranosu s M. semitendinosus M. biceps femoris M. tibialis anterior	M. quadriceps femoris M. semitendinosus M. semimembranosus M. biceps femoris M. tibialis anterior

	gluteus maximus Ischiocrurale Muskulatur	M. gluteus maximus M. adductor Magnus M. tensor fascia latae M. tibialis posterior M. peroneus longus		hallucis longus M. tibialis posterior M. peroneus longus M. peroneus brevis		iliacus M. tibialis anterior		
Functions	• heel contact to the ground	• shock absorption in knee and ankle joint • load transmission and stability in the hip • forward motion by heel rocke	• controlled forward motion of the tibia • shifting of the gravity centre to the front by ankle rocker	• controlled dorsal extension at the ankle joint with lifting the heel from the ground	• passive knee joint flexion of 40° • plantar flexion of the ankle joint	• min. 55° knee flexion for sufficient ground clearance	• increasing hip flexion to 25° • dorsal extension of the ankle joint to neutral-zero-position	• knee joint extension to neutral-flexion • preparation for stance phase

Appendix B

Statistical analysis

Normal walking

Table (2) Total variance of PCA (represent KMO values above 0.5)

Component	Initial Eigenvalues			Extraction Sums o Loadings f Squared		
	Total	Variance of%	Cumulative %	Total	Variance of%	**Cumulative %**
1	5.101	39.240	39.240	5.101	39.240	**39.240**
2	2.972	22.858	62.098	2.972	22.858	**62.098**
3	1.684	12.952	75.049	1.684	12.952	**75.049**
4	1.013	7.789	82.839	1.013	7.789	**82.839**
5	0.684	5.262	88.101	0.684	5.262	**88.101**
6	0.531	4.082	92.183	0.531	4.082	**92.183**
7	0.430	3.308	95.491	0.430	3.308	**95.491**
8	0.209	1.608	97.099	0.209	1.608	**97.099**
9	0.147	1.133	98.232	0.147	1.133	**98.232**
10	0.080	0.614	98.846	0.080	0.614	**98.846**
11	0.058	0.448	99.293	0.058	0.448	**99.293**
12	0.049	0.376	99.669	0.049	0.376	**99.669**
13	**0.043**	**0.331**	**100.000**	**0.043**	**0.331**	**100.000**

Component matrix

Table (3) Component matrix taken from PCA

Muscle	Component number												
	1	2	3	4	5	6	7	8	9	10	11	12	13
semimem	0.240	0.096	0.809	-0.156	0.429	-0.132	-0.190	0.013	-0.090	-0.054	-0.064	-0.024	**-0.023**
bifem	0.617	0.695	-0.102	0.051	0.142	-0.228	-0.136	-0.028	-0.034	0.091	0.059	0.128	**0.056**
vasmed	-0.147	0.553	-0.187	-0.620	0.223	0.445	0.005	-0.056	0.006	-0.015	0.039	0.011	**-0.002**
vaslat	0.439	0.830	-0.131	0.066	0.054	-0.171	-0.082	-0.009	0.199	-0.019	0.038	-0.130	**-0.002**
Rf	0.461	0.600	-0.096	0.369	0.240	0.092	0.454	0.011	-0.075	-0.013	-0.062	0.001	**-0.022**
medgas	-0.892	0.214	0.138	0.273	0.073	0.144	-0.047	0.058	-0.068	-0.010	0.012	-0.051	**0.159**
latgas	-0.888	0.160	0.242	0.251	0.091	0.052	-0.021	-0.076	-0.078	0.143	0.099	-0.040	**-0.091**
Tfl	-0.268	**0.566**	**0.512**	-0.241	-0.437	-0.118	0.165	-0.231	-0.027	0.001	-0.036	0.000	**0.025**
tibant	**0.813**	0.342	0.095	0.034	-0.341	0.103	-0.082	0.156	-0.196	-0.078	0.088	-0.029	**-0.025**
peronl	**-0.611**	**0.693**	-0.024	-0.003	-0.197	0.073	-0.157	0.230	0.036	0.067	-0.124	0.029	**-0.042**
soleus	**-0.871**	0.244	0.297	0.146	0.030	-0.057	0.088	0.074	0.122	-0.156	0.081	0.089	**-0.032**
Addmagnus	**0.587**	-0.276	**0.653**	-0.159	-0.036	0.105	0.221	0.186	0.129	0.110	0.046	0.001	**0.033**
gmax	0.665	**-0.010**	**0.305**	0.466	**-0.088**	**0.397**	**-0.213**	**-0.157**	**0.101**	**-0.012**	**-0.028**	**0.036**	**-0.007**

Running

Table (4) Total variance of PCA

Component	Initial Eigenvalues			Extraction Sums of Squared Loadings		
	Total	% of Variance	Cumulative %	Total	% of Variance	**Cumulative %**
1	4.501	34.620	34.620	4.501	34.620	**34.620**
2	2.755	21.191	55.812	2.755	21.191	**55.812**
3	1.842	14.169	69.981	1.842	14.169	**69.981**
4	1.111	8.548	78.528	1.111	8.548	**78.528**
5	0.781	6.009	84.537	0.781	6.009	**84.537**
6	0.729	5.608	90.145	0.729	5.608	**90.145**
7	0.551	4.242	94.387	0.551	4.242	**94.387**
8	0.248	1.907	96.294	0.248	1.907	**96.294**
9	0.176	1.355	97.649	0.176	1.355	**97.649**
10	0.136	1.043	98.691	0.136	1.043	**98.691**
11	0.097	0.745	99.437	0.097	0.745	**99.437**
12	0.042	0.326	99.763	0.042	0.326	**99.763**
13	**0.031**	**0.237**	**100.000**	**0.031**	**0.237**	**100.000**

Component matrix

Table (5) Component matrix taken from PCA

muscle	Component												
	1	2	3	4	5	6	7	8	9	10	11	12	**13**
semimem	-0.207	0.128	0.829	-0.025	0.098	0.347	-0.277	-0.068	0.143	-0.134	0.048	0.000	**-0.010**
bifem	0.675	0.371	0.130	-0.362	-0.250	0.336	-0.171	0.060	-0.013	0.199	-0.092	0.040	**-0.004**
vasmed	-0.189	-0.282	-0.523	0.425	0.101	0.628	0.105	0.110	0.041	0.026	0.029	-0.017	**0.004**
vaslat	0.585	0.691	-0.068	-0.217	0.166	0.133	-0.012	0.040	-0.226	-0.058	0.148	-0.062	**0.015**
rf	0.506	0.464	-0.286	0.338	0.257	-0.252	-0.378	0.210	0.117	0.012	0.019	0.033	**-0.003**
medgas	-0.847	0.404	-0.036	0.095	-0.248	0.001	0.037	-0.006	-0.050	0.038	0.155	0.131	**-0.023**
latgas	-0.874	0.361	0.186	0.119	-0.143	-0.043	-0.069	0.077	0.000	0.057	-0.021	-0.047	**0.137**
tfl	0.217	0.822	-0.170	0.338	0.142	0.091	0.080	-0.272	-0.041	-0.065	-0.128	0.048	**0.021**
tibant	0.873	0.170	0.089	0.162	-0.147	-0.073	0.220	-0.157	0.208	0.111	0.134	-0.035	**0.025**
peronl	-0.423	0.520	0.241	-0.296	0.457	-0.001	0.398	0.148	0.107	0.052	-0.031	0.016	**-0.015**
soleus	-0.775	0.547	-0.011	0.203	-0.107	-0.049	-0.102	-0.057	-0.001	0.103	-0.021	-0.110	**-0.096**
addmagnus	0.163	-0.445	0.646	0.431	0.335	-0.040	-0.014	-0.035	-0.168	0.167	0.016	0.019	**0.000**
gmax	**0.522**	**0.248**	**0.479**	**0.399**	**-0.385**	**-0.022**	**0.248**	**0.226**	**-0.045**	**-0.107**	**-0.052**	**-0.004**	**-0.023**

Appendix C

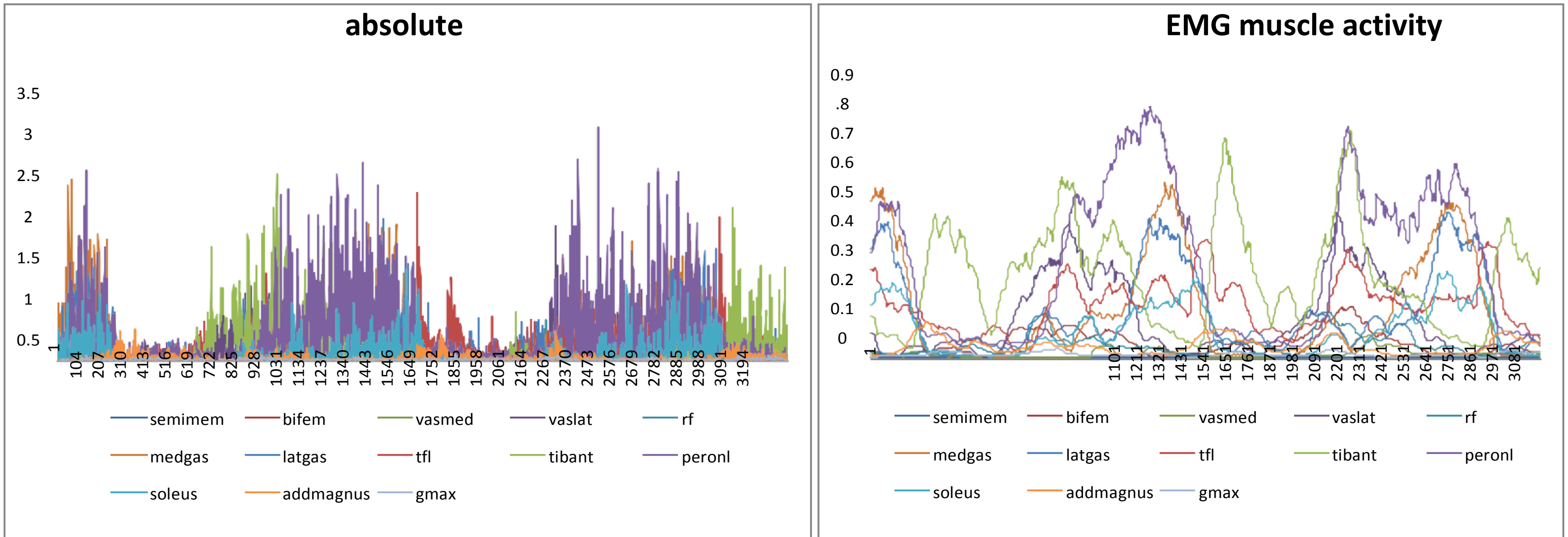


Figure 7 EMG data taken from the 13 muscles before(left) and after(right) moving average