Detailed Investigation of Knee Biomechanics during Posture Maintenance while Applying Different Static Loadings on the Spine

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Abstract: *The aim of this paper was to investigate in detail the biomechanics of the knee during different static loadings on the spine using electromyographic (EMG) signals from six main surface muscles acting in the knee joint; three components of the ground reaction force measured by a force plate; knee flexion joint angle measured by a flexible goniometer; and the distances between the bones (femur and tibia) forming the knee joint measured by an echograph. The measurements were taken without weight (reference straight position) and with a weight of 2, 5, 10, 15, 17, and 20 kg placed in a rucksack on the spine. The results showed that the forces in the horizontal and sagittal planes were negligible, and the reaction in the frontal plane increased and was linearly dependent on the carrying weight. The distance between bones decreased linearly with increasing weight for all participants from 3.94% to 53.92% from the referent position. The knee angle varied and in many cases decreased with increasing weight. The calculated correlation coefficients between mean EMG signals and loading weight showed that the adjustment of different subjects' musculature to increasing load is individual. In general, knee joint balance is a dynamic individual process.*

Keywords: *Lower limb, Knee, EMG, Force plate, Echograph, Static loading.*

Introduction

The knee joint has two degrees of freedom (DoF), flexion/extension and internal/external rotation. Ten muscles are responsible for these motions: 6 perform flexion, 2 perform extension, 5 perform internal rotation, and one performs external rotation. Knee stability can be investigated using one or two DoF models [9]. Since we investigated pose but not walking, we used only one DoF model, measuring only flexion/extension motion. According to the web page https://www.cdc.gov/ncbddd/jointrom/index.html, the range of knee joint motions is as follows: knee flexion, 152.6° (151.2° ÷ 154.0°) for females and 147.8° (146.6°÷ 149.0°) for males; knee extension, 5.4° (3.9°÷ 6.9°) for females and 1.6° (0.9°÷ 2.3°) for males.

Static investigations of knee joint stability are often focused on stretching [7, 15, 18], yoga poses (http://www.doyoga.com/articles_all/7_July_07_knees.pdf) [2, 11, 20], and isometric back squats [12, 19]. Electromyographic (EMG) signals are often used for the estimation of muscle activity in normal and pathological conditions [1, 4, 8, 21].

Echography (ultrasonography, sonography) is often used to diagnose different muscle, tendon, and joint injuries [13, 16, 22]. This method is rarely used for measuring bone lengths or bone-to-bone distances.

Using a combination of muscle EMG, echography (measurement of the distance between the bones forming the joint), and joint flexion/extension angle and ground reaction force for the investigation of knee joint stability has not been reported in the literature. This combination can give a general impression of the stability of the knee joint during different static loadings. What happens inside the joint, what is the behaviour of the synovial liquid, and how the friction is changed, depends on the forces acting in the joint, i.e., muscle forces, reaction forces, and weight forces. To make a model of the knee joint using fluid dynamics all these forces have to be calculated. The distance between bones is a very important quantity in such a model and it is measurable using echography. Data from tensometric platforms can be used in dynamic conditions, but sonography methods cannot. Using EMG to determine the synchronization between different synergistic and antagonistic muscles during dynamic conditions is also controversial. Therefore, we chose to investigate static knee loading through different loads placed on the spine. The aim of the paper is to investigate in detail the biomechanics of the knee during the loading of the spine with different external weights by using EMG signals from the six main surface muscles acting on the knee joint, and measuring ground reaction forces by the use of a force plate, knee flexion joint angle by the use a flexible goniometer, and the distances between the bones (femur and tibia) forming the knee joint by the use of the echography.

Materials and methods

Experimental equipment and experimental procedure

The experiments were performed on 25 healthy subjects, identified as SUB1, SUB2, …, SUB25. The subjects were athletic students from the National Sports Academy "Vassil Levski", Sofia, Bulgaria. They did not report any health problems. They completed an injury record and were informed in detail about the aim of the experiments and the procedure. All participants gave informed consent. The mean height (± SD) of the subjects was 180.5 ± 2.5 cm, and the mean weight was 88.9 ± 1.6 kg. The mean age of the subjects was 25 ± 5 years, and all subjects were right-handed. The experimental procedure was approved by the Scientific Council of the Institute of Biophysics and Biomedical Engineering, Sofia, Bulgaria.

The subject stood on a tensometric platform (Fig. 1) with a rucksack on the spine. The components of the ground reaction forces, $\mathbf{R_x}$, $\mathbf{R_y}$, and $\mathbf{R_z}$, were measured and stored on a hard disk for further processing. The EMG signals from the six main knee surface muscles (*m. gastrocnemius lateralis* [**GAL**], *m. gastrocnemius medialis* [**GAM**], *m. rectus femoris* [**RF**], *m. vastus lateralis* [**VL**], *m. biceps femoris* [**BF**], and *m. semitendinosus* [**ST**]) and flexion/extension knee angle were measured using the 8-channel telemetric system Telemyo 2400G2 from Noraxon, Inc. with online monitoring, and the experimental data were saved for further offline processing. The reference electrode was placed at the caput fibulae. The surface EMG signals were recorded by "Skintact-premier" F-301 Ag/AgCl disk electrodes (Fig. 1), which had a 9 mm diameter, and additional Hellige EMG conductive gel was used for better conductance. The pairs of electrodes were placed at a 3 cm center-to-center distance on the skin of the subjects, which was first cleaned with alcohol and dried. The electrode locations were determined according to international EMG guidance (SENIAM project, http://www.seniam.org/). The electrodes were oriented parallel to the muscle fibers. The sampling frequency was 1500 Hz, and the duration of each motor task was one-half of a minute. The flexion/extension angle was measured by a flexible goniometer from Noraxon. All measurements were taken on the right leg. The experimental data for EMG signals and angle were stored on a hard disk as an ASCII file and were further processed by in-house MATLAB program.

Fig. 1 Experimental setup

The tensometric platform on which the subjects stood was a Bertec model (weight 600 mm, length 400 mm, height 50 mm) (Fig. 1), and the data were sampled at a frequency of 1 kHz. The software for collecting and storing the data was a Noraxon system using the module "MyoForce". The ground reaction force and its components were smooth, without artefacts, so mean values in the chosen time interval were calculated as the value $\mathbf{R} = \sqrt{\mathbf{R_x}^2 + \mathbf{R_y}^2 + \mathbf{R_z}^2}$ using the mean values of the components of the reaction.

The experimental procedure was as follows. The first motor task was performed without additional load. The subject stood on two legs on the force platform and was asked to stay upright for half a minute without moving. The measurements (EMGs, joint angle, and ground reaction forces) were synchronized by a sound signal from a computer and a synchronization LED for a second synchronization check. After two minutes of rest, an additional load of 2 kg was placed in the rucksack, and another half of minute measurement was taken. The procedure was repeated with the following weights: 5, 10, 15, 17, and 20 kg.

During quiet standing, the distance between the tibia and femur bones forming the knee joint was measured using ultrasound scanning (Figc. 2 and 3) simultaneously with the previously described EMG and tensometric data. The portable ultrasound system used was a Vinno 6, China, 8 MHz transducer frequency. The obtained pictures in DICOM format were analyzed with RadiAnt DICOM Viewer, version 2022.1.1. (64-bit). Statistical analysis of ultrasound data was conducted with the program SigmaPlot 10.

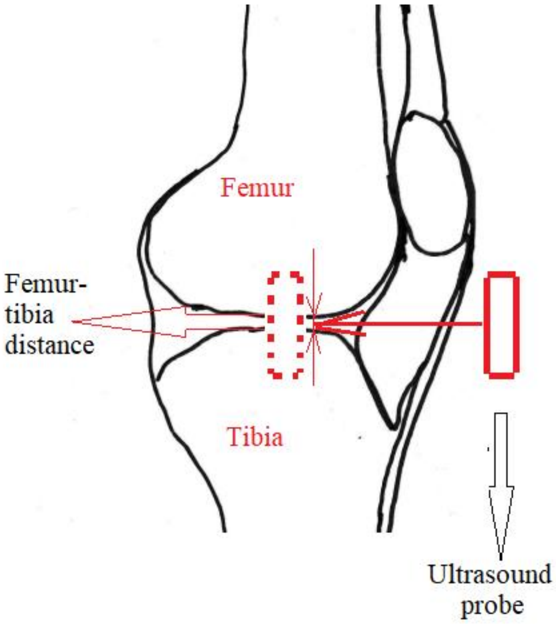
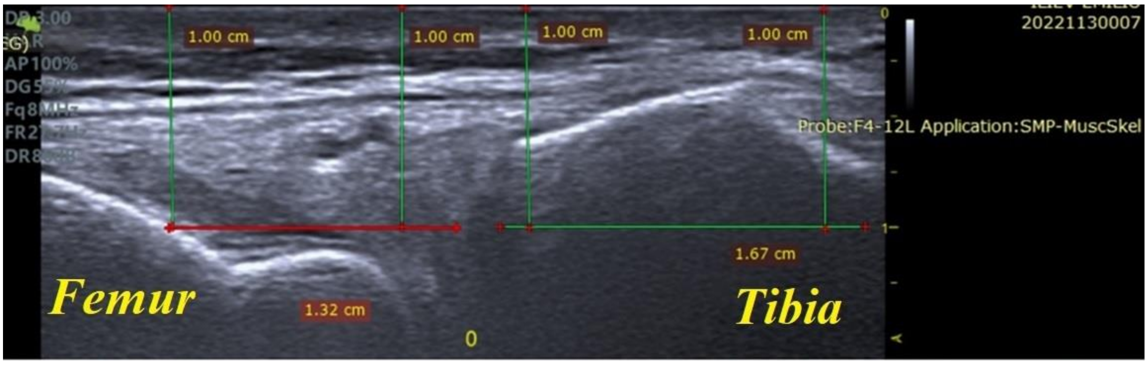
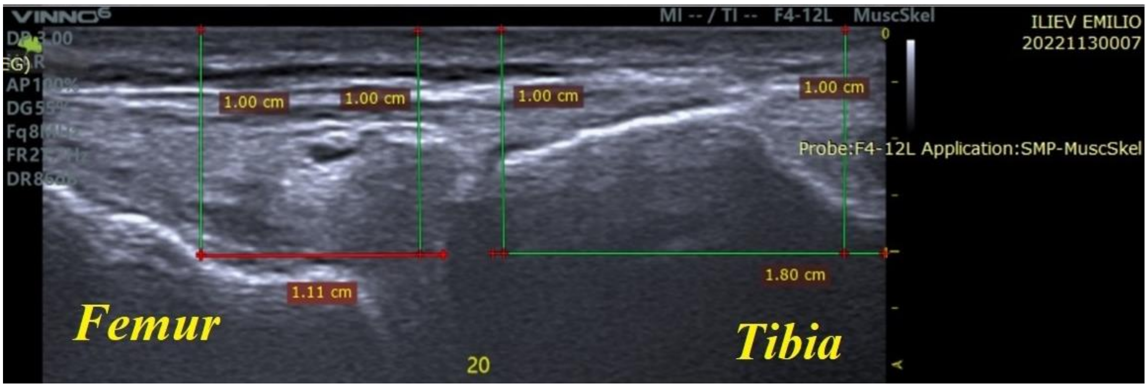


Fig. 2A The anatomic schemes for two bone distance measurements with ultrasound transducer (probe) position



without loading



with loading of 20 kg

Fig. 2B The screen view of the echograph with measured distances between the femur and tibia bones in the knee joint for one participant (SUB11)

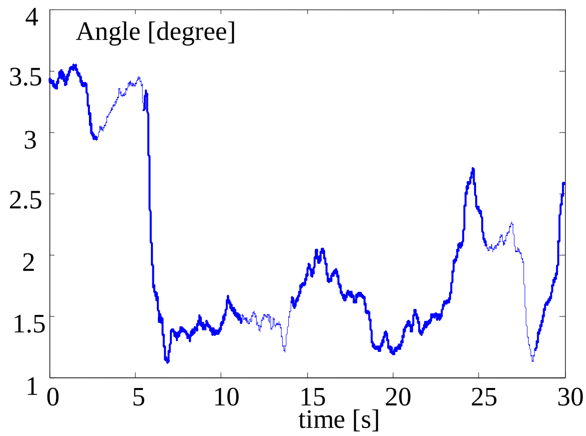


Fig. 3 Ankle flexion/extension angle for SUB1 for the experiment with 5 kg loading

Data processing

All the data from the EMG and force plate were processed by in-house MATLAB programs. The processing for EMG signals consisted of Butterworth high-pass (cut-off 20 Hz, order 4) and low-pass (cut-off 300 Hz, order 4) filtering in a visually chosen suitable time interval, rectifying, smoothing (10 samples) and calculating the mean value [14]. The recording time for each experiment lasted 30 seconds, but a time interval of a minimum of 10 ms was visually chosen, aiming to have a smooth, nearly constant angle and no artefacts in all EMG signals. The angle was sufficiently smooth, so additional processing was not necessary. The mean value was calculated in the chosen time interval. The same procedure was used for $\mathbf{R_x}$, $\mathbf{R_y}$ $\mathbf{R_z}$, and **R**.

For each subject, an Excel table was prepared, as shown in Table 1 for SUB2. This table has 12 columns and they are subjects of further statistical analysis.

Table 1. Measured and processed parameters. Mean EMG signals in arbitrary units for all 6 muscles; knee angle in degrees, joint reaction, and its components.

Weight, [kg]	**EMG [a.u.] muscles**						**Angle, [deg]**	**Reaction, [N]**			
	GAL	**GAM**	**RF**	**VL**	**BF**	**ST**		$\mathbf{R_x}$	$\mathbf{R_y}$	$\mathbf{R_z}$	**R**
0	0.0441	0.0877	0.0079	0.0209	0.1243	0.0199	0.3382	1.8418	2.1916	775.4539	775.4591
2	0.293	0.0250	0.0066	0.0055	0.0582	0.0133	0.7847	0.5755	3.9962	796.5978	796.608
5	0.0324	0.0260	0.0070	0.0055	0.0115	0.0066	2.2590	0.5000	3.0000	800.0000	810.0000
10	0.0278	0.0750	0.0072	0.0051	0.0294	0.0086	2.6636	2.5341	3.779	873.2724	873.2843
15	0.0194	0.1735	0.0079	0.0071	0.0074	0.0071	1.9046	0.4173	0.7318	922.0601	922.0605
17	0.0142	0.0823	0.0064	0.0062	0.0312	0.0099	1.2742	1.2406	2.9461	941.0779	941.0833
20	0.0312	0.0218	0.0232	0.0126	0.0144	0.0204	0.0148	-0.6028	4.9986	971.0623	971.0753

Correlation analysis was performed. Correlation coefficients between column 1 (loading); columns with EMG signals of the 6 muscles; angle column; and columns with the reactions $\mathbf{R_x}$, $\mathbf{R_y}$, $\mathbf{R_z}$, and **R** were calculated.

The distance D between the femur and tibia for all participants at the reference position and different loadings was measured in centimetres (Fig. 2B). For four of the subjects, the ultrasound scan was low quality because of knee movement, ultrasound transducer position changes around the knee joint centre. Therefore, these subjects were excluded from further statistical analyses of bone-to-bone distance. Regression equations were obtained.

Results

In Fig. 3, an example of knee angle fluctuation is shown. In general, the range (absolute value, approximately 5.8 degrees) and the fluctuations increased with increased carrying weight in the direction of flexion. Hence, the space between the tibia and femur decreased. The instructions to the subjects were to stay upright and stable without movement, but small oscillations of the body were observed for each subject (Fig. 3).

In Figs. 4A-C, the successive steps of EMG processing are illustrated. In Fig. 4A, the original data are given. In Fig. 4B, the filtered signals are shown, and in Fig. 4C, the last stage, the smoothed and rectified EMG signal in a visually chosen time interval without artefacts is shown.

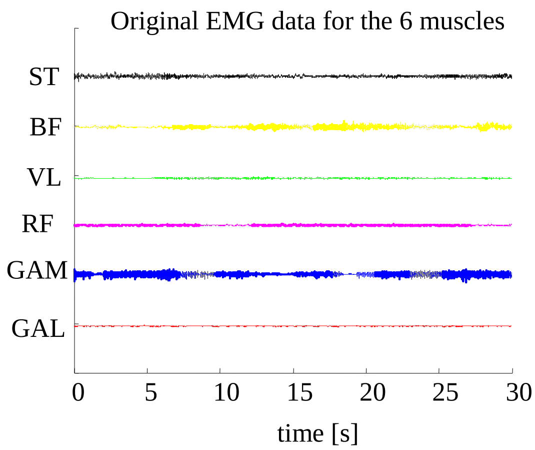


Fig. 4A Original EMG signals for the 6 muscles during the motor task with a 5 kg load for SUB1

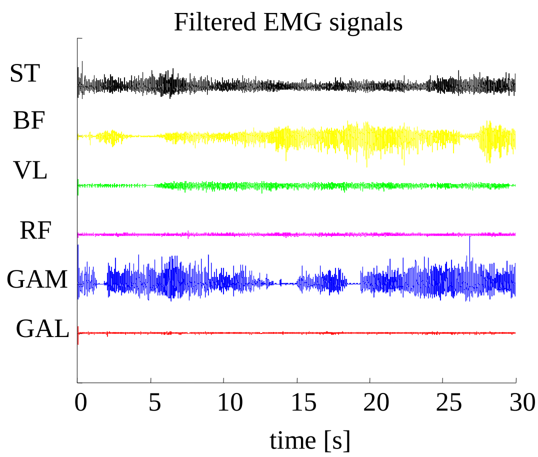


Fig. 4B The filtered and smoothed EMG signals for the 6 muscles during the motor task with a 5 kg load for SUB1

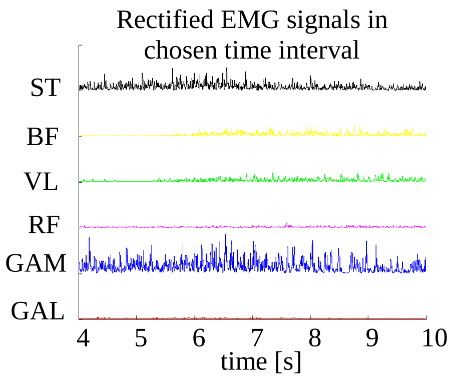


Fig. 4C The processed EMG signals (filtered, rectified and smoothed) for the 6 muscles during the motor task with a 5 kg load for the chosen time interval for SUB1

Fig. 5 shows the ground reaction force **R** and its components $\mathbf{R_x}$, $\mathbf{R_y}$, and $\mathbf{R_z}$ within the chosen time interval during the motor task with a 5 kg load for the chosen time interval for SUB1.

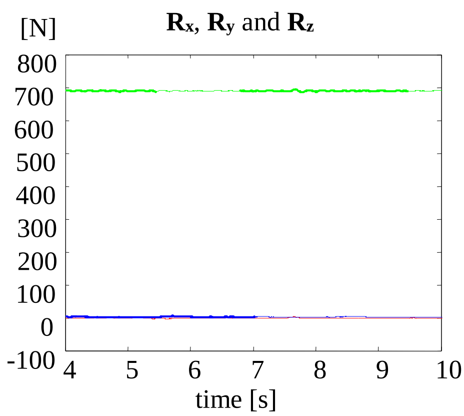


Fig. 5 The force data for SUB1 with a 5 kg load for the chosen time interval. $\mathbf{R_x}$ – red, $\mathbf{R_y}$ – blue, $\mathbf{R_z}$ – green. **R** (not shown) and $\mathbf{R_z}$ nearly coincide, and $\mathbf{R_x}$ and $\mathbf{R_y}$ nearly coincide and are approximately zero.

The correlation coefficients (CorCo) were calculated between all parameters for each subject, except the distance between the tibia and fibula, which is analysed separately below. In Table 2 these correlation coefficients are shown for SUB12. The correlation coefficients in diagonal elements are ones. The CorCo between **R** and $\mathbf{R_z}$ and the applied weight are logically close to one, indicating that there are linear dependences (positive) between these two parameters and the loading. In contrast, $\mathbf{R_x}$ and $\mathbf{R_y}$ do not change in synchronization with the applied loading. The remaining CorCo for (i-j) ($i \neq 1$, $j \neq 1$ and $i \neq j$) are mixed, positive, negative, up or down than 0.5.

To see the entire picture, in Table 3, all CorCo between the first column and the other 11 parameters for all investigated subjects are shown. The CorCo between loading and **R** and $\mathbf{R_z}$ are always very close to one. It is also clear that for the remaining parameters, there are no trends that were observed for all subjects. Hence, postural stability is intrasubjective.

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Table 2. Correlation coefficients between the columns of Table 1 and the measured parameters* for SUB12

(1-1)	(1-2)	(1-3)	(1-4)	(1-5)	(1-6)	(1-7)	(1-8)	(1-9)	(1-10)	(1-11)	(1-12)
1.0000	-0.6705	0.2083	0.5596	-0.2438	-0.6914	-0.0638	0.0287	-0.4111	0.1321	0.9934	0.9976
	(2-2)	(2-3)	(2-4)	(2-5)	(2-6)	(2-7)	(2-8)	(2-9)	(2-10)	(2-11)	(2-12)
	1.0000	-0.3799	0.1721	0.7077	0.6655	0.6053	0.3674	0.1211	0.1836	-0.6669	-0.6696
		(3-3)	(3-4)	(3-5)	(3-6)	(3-7)	(3-8)	(3-9)	(3-10)	(3-11)	(3-12)
		1.0000	-0.3364	0.0517	-0.0280	-0.3374	-0.2961	0.2496	-0.8695	0.2399	0.2265
			(4-4)	(4-5)	(4-6)	(4-7)	(4-8)	(4-9)	(4-10)	(4-11)	(4-12)
			1.0000	0.3267	-0.2419	0.6322	0.5827	-0.6400	0.5541	0.5605	0.5618
				(5-5)	(5-6)	(5-7)	(5-8)	(5-9)	(5-10)	(5-11)	(5-12)
				1.0000	0.7437	0.8169	0.6831	0.0530	-0.0928	-0.2061	-0.2226
					(6-6)	(6-7)	(6-8)	(6-9)	(6-10)	(6-11)	(6-12)
					1.0000	0.5871	0.4756	0.4654	-0.1022	-0.6308	-0.6566
						(7-7)	(7-8)	(7-9)	(7-10)	(7-11)	(7-12)
						1.0000	0.9132	-0.2049	0.4189	-0.0103	-0.0313
							(8-8)	(8-9)	(8-10)	(8-11)	(8-12)
							1.0000	-0.4451	0.3234	0.0779	0.0591
								(9-9)	(9-10)	(9-11)	(9-12)
								1.0000	-0.2060	-0.3772	-0.3928
									(10-10)	(10-11)	(10-12)
									1.0000	0.1338	0.1348
										(11-11)	(11-12)
										1.0000	0.9990
											(12-12)
											1.0000

* The number of correlated columns is given in brackets; the means of the columns are as follows: 1 – weight; 2 – **GAL**, 3 – **GAM**, 4 – **RF**, 5 – **VL**, 6 – **BF**, 7 – **ST**, 8 – angle, 9 – $\mathbf{R_x}$, 10 – $\mathbf{R_y}$, 11 – $\mathbf{R_z}$, 12 – **R**.
Red indicates coefficients with absolute values higher than 0.5.

The calculated correlation coefficients showed versatile results. As to the muscle activity, some correlation coefficients are positive, and some are negative. This means that while increasing loading sometimes the activity of the same muscle can increase or decrease. The muscle GAL and GAM show for most of the subjects (but not for all) positive CorCo bigger than 0.5. Surprisingly the activity of the muscle BF decreases with increasing the load. The remaining muscles show varied coefficients.

That is why we plotted the figures showing the dependence of mean EMG signals on applied weight for all muscles. Two examples are shown in Figs. 6A and 6B. Only two EMG signals showed a tendency to increase with increasing load, i.e., **GAM** (Fig. 6A) and **RF**. Less electrical activity was observed in **GAL** and **BF**.

Table 3. Correlation coefficients between the columns of Table 1 and the measured parameters for all subjects and loading weights*

	(1-1)	(1-2)	(1-3)	(1-4)	(1-5)	(1-6)	(1-7)	(1-8)	(1-9)	(1-10)	(1-11)	(1-12)
SUB1	1.0000	0.6685	-0.5163	0.6874	0.5966	-0.6318	0.0174	-0.4165	0.0194	0.0734	0.9999	0.9999
SUB2	1.0000	0.8580	0.8879	0.1554	0.5525	0.1335	-0.0493	-0.8950	0.6734	-0.1835	0.9998	0.9998
SUB3	1.0000	0.6733	0.7692	-0.6600	-0.7289	-0.5956	0.5153	-0.9213	-0.8686	0.0458	0.9999	0.9999
SUB4	1.0000	0.3505	-0.5291	0.5745	0.6932	0.6679	0.4557	0.4840	0.5410	0.6300	0.9999	0.9999
SUB5	1.0000	0.7538	0.8207	0.8817	0.2874	-0.0329	0.5928	0.4227	-0.7169	0.4066	0.9793	0.9793
SUB6	1.0000	0.9196	0.8573	-0.6640	-0.7376	-0.3191	-0.8692	-0.0703	0.1657	0.6552	0.9998	0.9998
SUB7	1.0000	0.8944	0.8749	-0.1015	-0.0718	-0.2219	0.4204	-0.9554	0.5040	0.3635	0.9999	0.9999
SUB8	1.0000	0.9601	0.8590	-0.4479	-0.1267	-0.6826	-0.7803	-0.3723	0.3523	-0.1845	0.9998	0.9998
SUB9	1.0000	0.3441	0.2196	0.4005	0.3084	-0.4253	0.0530	0.0289	0.3419	0.4242	1.0000	1.0000
SUB10	1.0000	0.6879	0.6717	0.5964	0.5823	0.5756	0.3114	-0.3523	0.6417	0.4353	0.9999	0.9999
SUB11	1.0000	0.7099	0.4073	0.4857	0.7875	-0.8897	-0.5629	0.0662	0.9999	-0.1389	0.9444	0.9999
SUB12	1.0000	-0.6705	0.2083	0.5596	-0.2438	-0.6914	-0.0638	0.0287	-0.4111	0.1321	0.9934	0.9976
SUB13	1.0000	0.4914	0.1631	0.1224	0.3469	0.8995	0.6643	-0.3086	0.6097	0.3864	0.9986	0.9986
SUB14	1.0000	0.9163	0.8063	-0.5771	-0.4909	-0.0287	-0.1891	-0.9133	0.2591	0.3321	1.0000	1.0000
SUB15	1.0000	0.7081	0.5587	0.7864	0.8986	0.9264	0.8604	-0.7934	-0.3748	0.6244	1.0000	1.0000
SUB16	1.0000	0.4138	0.9244	0.4329	0.5983	-0.4458	-0.4058	-0.8542	0.8777	0.0512	0.9999	0.9999
SUB17	1.0000	0.3636	0.6148	0.6850	0.1039	0.1368	-0.3546	-0.7396	0.3938	-0.8993	0.9986	0.9986
SUB18	1.0000	0.0478	0.1006	0.4385	0.8492	-0.3170	-0.5386	-0.7872	0.7068	-0.1137	0.9999	0.9999
SUB19	1.0000	0.8924	0.0726	0.6874	0.8368	-0.4094	-0.4216	-0.1483	0.3890	0.4602	0.9999	0.9999
SUB20	1.0000	0.2946	0.4215	-0.2938	-0.2622	-0.6639	-0.5483	-0.5649	0.9544	-0.4904	0.9999	0.9999
SUB21	1.0000	0.8231	0.7033	0.8826	0.5777	-0.1379	-0.2181	-0.2621	0.8211	0.6112	0.9998	0.9998
SUB22	1.0000	0.9119	0.8562	0.4253	0.3091	-0.4061	-0.4568	-0.9137	0.4374	-0.3343	0.9999	0.9999
SUB23	1.0000	0.8707	0.8063	0.8910	0.5856	0.5317	0.7870	-0.4936	0.6918	0.1897	0.9815	0.9895
SUB24	1.0000	0.2174	0.0928	0.3164	-0.5013	-0.7332	-0.7546	0.5379	0.9999	0.1989	0.9391	0.9999
SUB25	1.0000	0.5189	0.4170	0.7773	0.7581	-0.6592	0.2010	-0.7242	-0.3637	-0.5423	1.0000	1.0000

* Red indicates coefficients with absolute values higher than 0.5.

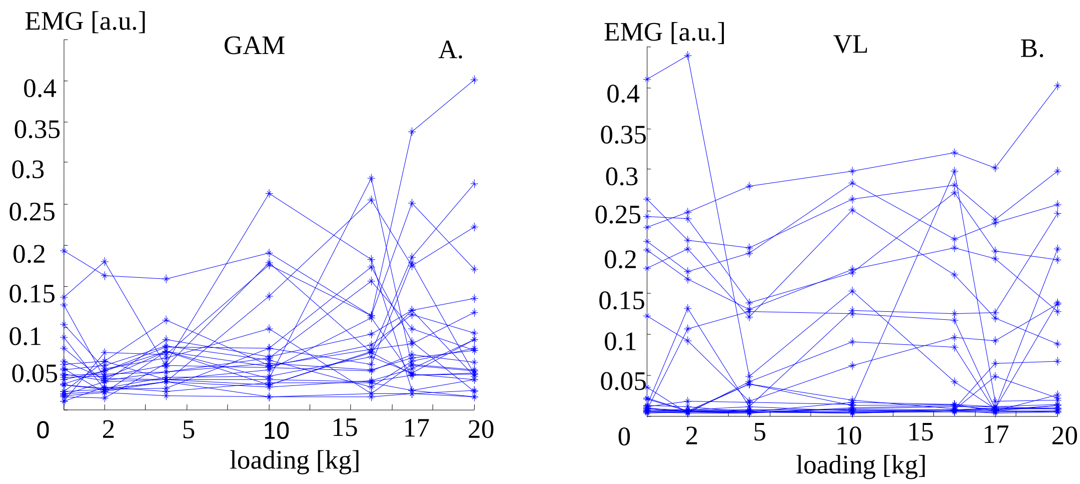

Fig. 6 The mean values of the processed EMG signals of two muscles: A) GAM; and B) VL for all subjects for all loading weights.

It has been suggested that the gastrocnemius muscle is largely responsible for the phase control of anterior-posterior balancing during standing [3]. Both young and people activate the lateral gastrocnemius for less than 20% of the total standing time [5]. Regarding the activity of another multiheaded muscle acting on the knee joint during standing, that is the quadriceps muscle, Seng-Jung et al. [17] reported the influence of foot position on the activity of its individual muscles. The activity of the vastus lateralis, rectus femoris, and vastus medialis was greatest when the foot was rotated outwards.

Predicted values for femur-tibia distances D for the seven tested loads were determined from the individual regression equations for all subjects. All collected data fell within the 95% confidence interval around the regression line. The relative reduction in femur-tibia distances in percentage was calculated based on the obtained regression model (Table 4).

The depth of the ultrasound measurement depended on the individual participant. For example in larger (heavier) subjects, the knee joint is located further inwards relative to the forehead of the ultrasound probe (Fig. 2B). Therefore, the depth at which the femur and tibia surfaces are visualized was greater in participants SUB4 and SUB5 (Table 4).

The depth of the ultrasound measurement in this study influenced the quality of the obtained ultrasound pictures because the intensity of the sound waves emitted from the transducer decreases with the depth of scanning.

Table 4. Participant data for L – depth under ultrasound transducer for measured distances; D – femur-tibia knee distance; α – angle coefficients of linear regression equations; and ΔD – relative reduction in femur-tibia distances in percentage based on the obtained regression model. NA – not available.

Subjects	**L, mm depth**	**D, cm 0 kg**	**D, cm 2 kg**	**D, cm 5 kg**	**D, cm 10 kg**	**D, cm 15 kg**	**D, cm 17 kg**	**D, cm 20 kg**	α	**ΔD, %**
SUB1	1.0000	1.5000	1.5800	1.3000	1.3900	1.2600	1.2300	1.3600	-0.0117	15.7400
SUB2	1.0000	1.6500	1.5700	1.6000	1.5500	1.4600	1.7000	1.3700	-0.0072	8.8100
SUB3	NA	2.0500	2.0000	1.8500	NA	1.8000	1.7800	1.7100	-0.0147	14.6100
SUB4	1.2500	1.9300	1.8200	NA	1.7600	NA	1.5400	1.4100	-0.0233	24.2700
SUB5	0.8750	1.0900	1.0800	1.0700	1.0300	0.9790	NA	0.9900	-0.0059	10.7370
SUB6	1.5000	2.2100	1.9000	2.0500	2.0600	1.5300	1.4600	1.6000	-0.0327	30.3500
SUB7	1.5000	0.595	NA	NA	NA	NA	NA	0.4890	-0.0053	17.8100
SUB8	1.0000	1.2500	1.2200	1.2000	1.2500	1.2100	NA	1.1800	-0.0022	3.5400
SUB9	1.0000	1.4500	1.0200	1.0900	1.1600	0.9960	1.0300	1.0500	-0.0119	19.1800
SUB10	1.0000	0.8950	0.6510	NA	0.6730	0.4560	0.4390	NA	-0.0220	53.9200
SUB11	1.0000	1.8600	1.8000	1.7700	1.7500	1.7700	NA	NA	-0.0052	5.7000
SUB12	1.0000	1.2800	1.2400	1.1400	1.1300	1.0900	1.0600	1.0200	-0.0116	18.5500
SUB13	1.1250	1.8400	1.8000	1.8900	1.6600	1.6100	NA	1.6100	-0.0139	15.000
SUB14	1.0000	1.7800	1.5400	1.5200	1.4700	1.6200	1.7100	1.4900	-0.0032	3.9400
SUB15	1.2500	1.6600	1.5800	1.5100	1.4700	1.5700	1.3800	1.3700	-0.0112	13.8900
SUB16	0.7500	1.6200	1.4400	1.5100	1.5200	NA	1.4400	NA	-0.0094	11.7600
SUB20	1.0000	1.8400	NA	NA	NA	NA	NA	1.3900	-0.0225	24.4600
SUB21	1.0000	1.4100	1.3600	1.4000	1.3200	1.2500	1.1900	1.1700	-0.0121	17.0800
SUB22	0.9000	1.8000	NA	1.7100	1.5700	1.5500	1.5600	1.5400	-0.0132	14.9000
SUB23	1.0000	2.1300	1.8700	2.0500	1.7500	1.6500	1.7100	1.5200	-0.0256	24.7800
SUB25	1.0000	1.1900	1.1800	1.1400	1.0500	1.0700	NA	1.0800	-0.0064	10.9200

The angle coefficients in Table 4 indicate that for most participants there were decreases in femur-tibia distances (a minus sign indicates a negative slope) based on the linear fitting of data for all participants in the study (angle coefficients were obtained from linear equation coefficients a_1) (see the caption of Fig. 7).

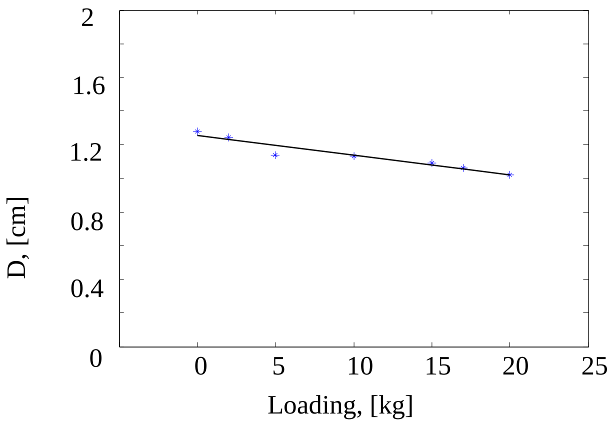

Fig. 7 Linear fitting of data for one participant (SUB3): '*' values of D for different loadings; black line – linear approximation ($y = a_1 * x + a_2$, $a_1 = -0.0116$, $a_2 = 1.2516$).

The sound waves emitted from the transducer are transmitted into the body, reflected off the tissue interface, and returned to the transducer (http://pie.med.utoronto.ca/OBAnesthesia/OBAnesthesia_content/OBA_ultrasonographyBasics_module.html). The increase in the personal femur-tibia distances in Table 4 was related to small participant movement (SUB20, SUB10) and/or ultrasound transducer movement regarding the participant's knee joint.

For each subject, a graph was made of the measured distances as a function of loading. A linear regression analysis was then performed to determine the analytical equation fitting the experimental data. This procedure allowed us to determine the angular coefficients for each participant separately (Fig. 7, Table 4).

Discussion

The main conclusion from our investigation is that balance in the knee joint is very complex and performed individually, especially according to muscle activity (Table 3). Logically, **R** and $\mathbf{R_z}$ were linearly proportional to the applied load. The load was placed in a rucksack on the spine nearly symmetrically, and the weights had a perpendicular weight force with nearly no components in the X and Y axes. That is, the CorCo in columns (1-9) and (1-10) are mixed. Some subjects bend slightly forward, some backwards, some to the right, and some to the left, depending on the position of the weight in the rucksack. From the visual inspection of the camera recordings, it could be summarized that although the participants were told to stand and maintain their initial body position and look ahead at a fixed point, this could not be achieved during the entire experimental recording. Even when the body appeared to be fixed and motionless, the arms were often swaying back and forth, which means that the body continuously swayed and was not static all the time. As the weight increased, some of the participants reacted with an increase in sacral lordosis or a forward tilt of the pelvis to oppose the increasing tension on the back. Others maintained the upright position of the body but moved forward along the longitudinal axis to maintain balance. Even during standing without the external load, the body sway from toes to heels was observed in the participants. Some participants pointed their fingers straight ahead, while others had them pointed slightly to the side. In dependence on individual characteristics of the pose, more or less different muscles of the lower limb were activated.

Surprisingly, the absolute values of the CorCos in columns (1-8) in Table 3 do not show values close to one. Our visual observation was that with increasing loading, the knee was flexed. However, the hip was also flexed, and it seemed to flex more than the knee. As shown in Fig. 3, there were fluctuations in the angle, and the interval chosen by the operator fell into different phases. Another problem is the normalization of the goniometer (zero offsets) and the place where its arms are fixed. A possible solution is using a smaller interval (less than 30 s) for the calculation of the mean joint angle.

In most scientific papers, EMG normalization is performed, initially recording the maximal isometric contractions. We did not use normalization for several reasons. First, it was not very clear in which positions should the maximal isometric force tested for the 6 chosen muscles. Second, the measured EMG signals were very low, and normalizing them to the maximal isometric values would have resulted in very low values. As we did not compare muscle EMG signals with muscle forces and did not draw conclusions on which muscles are more active, we decided only to correlate their activity.

Our observation is that the knee is a damper. The six observed muscles showed small activity, especially **ST**, **BF,** and **GAL**. All of them support the stability of the knee, and only the **RF** and **GAM** muscles showed some increase in activity with increasing load. From visual observation, it can be concluded, that the stability of the hip is also very important. Most of the subjects bent slightly forward with increasing load.

The contact surface between the femur and the tibia varies between 2 cm^2 and 6 cm^2 [10]. To the best of our knowledge, this study was the first to investigate the reduction of the distance between the femur and the tibia in the knee joint at 0° flexion with increased load. These results will be the basis for the following: (1) determination of the change in the contact area between the femur and the tibia under different axial loads; (2) evaluation of the deformation of the cartilage tissue from the contact area between the femur and the tibia under different axial loads; (3) modelling the interaction between cartilage deformation and interstitial fluid flow from the cartilage into the joint cavity under loading conditions.

The difficulties of the method for accurate measurement of the femur-tibia distance are related to the following. First, ensuring the immobility of the ultrasound transducer regarding the knee joint and second, minimizing small movements in the knee of the tested participants.

This investigation is important for sportspeople (weightlifters) and schoolchildren. It can be concluded that if the weight applied to the spine is symmetrical, the lower limb muscles do not increase muchly the activity, but the distance between the bones forming the knee joint decreases. This can increase the friction in the joint leading to deformations of the joint cartilage.

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