Human Locomotion Strategies Under Changed Bodyweight Support

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INTRODUCTION: The aim of this study was the analysis of human musculoskeletal system energy costs of normal walking and walking under reduced weight loading.

METHODS: There were 15 subjects who participated in the study. We analyzed the biomechanical parameters of walking under different musculoskeletal system loads. The subjects walked on a treadmill at a pace of 90 steps/min under various loading conditions: 1) 100% bodyweight loading, corresponding to the terrestrial surface; 2) 38% bodyweight loading, corresponding to the surface of Mars; and 3) 17% bodyweight loading, corresponding to the surface of the Moon. Joint angles and angular velocities were recorded from the hip, knee, and ankle.

RESULTS: We analyzed changes in joint phase trajectories and the ratio of kinetic extension energy to kinetic flexion energy in the joints. We observed changes in kinetic energy parameters associated with both flexion and extension motions in the joints of the feet while walking under various loads. In terrestrial conditions (walking under 100% bodyweight), flexion kinetic energy in the hip joint prevailed over extension kinetic energy by 90%, with a small variation equal to 22%. If weight loading decreased up to 17% (lunar conditions), the difference between flexion and extension kinetic energies diminished, and eventually reached only 9%. The ratio of flexion energy and extension energy in the ankle joint equalized under lower loading conditions. Thus, 38% bodyweight loading was sufficient for approximation of flexion and extension energy values.

DISCUSSION: Our results revealed that phase trajectories shifted toward smaller joint angles and a decreased ratio between extension kinetic energy and flexion kinetic energy in the knee joint of all subjects. However, significant differences in the ratio of flexion and extension kinetic energy in the knee joint under bodyweight support were not found. The methods used for musculoskeletal system assessments that were proposed in our work can be used in clinical practice to evaluate the effectiveness of rehabilitation measures in a patient's musculoskeletal system disorders.

KEYWORDS: musculoskeletal system, bodyweight support, joint angles, angular velocities, phase trajectories, the energy cost of walking.

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Currently, more and more studies in space biology and medicine are devoted to modeling physiological changes in various body systems in terrestrial conditions and under lower gravity levels[1,4,21] to better understand changes in humans which occur during spaceflights to celestial bodies (e.g., the Moon and Mars). Previous studies of the musculoskeletal system following zero gravity exposure have shown changes throughout all parts of the motor system.[5,12,20] Studies performed in real and simulated microgravity environments showed a wide range of muscle changes (atony, atrophy) and sensory input changes, including changes in supporting,

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muscular, and vestibular inputs.[9–11] These changes adversely affected motor control performance[7,13] and disrupted postural regulation, precise motor control,[6,8,19] and locomotion.[2,18,25] We suggest that in the above gravity environments, such as the lunar surface, cosmonauts will experience similar changes in musculoskeletal function.

Studies on the motor activity and vertical stability under modeled reduced gravity showed that humans were able to maintain a vertical posture within the context of lunar gravitation, but posture was "slightly bent down." Such studies were performed in the USSR at the same time as various foreign studies.[3,15,16]

Studying the human motor system and, in particular, the musculoskeletal system, is not an easy task. First of all, given the complexity of human locomotion, one has to consider both internal and external factors. External biomechanical locomotion parameters include joint angles, angular velocities, and angular accelerations. They also include locomotion effort, amplitude, frequency, and speed of both individual body parts and the entire human body.[24] Internal locomotion parameters include muscle activity to ensure the performance of a specific movement (EMG-activity) and the work through which these movements manifest (EMG-cost).[14] The kinematic composition of step movements involves total kinematic control; for example, control of step time and step length are performed with precise control of the joint angles.

We studied human locomotion and its relationship to energy costs under normal walking and walking under reduced musculoskeletal loads. We hypothesized that vertical bodyweight supports changes in locomotion strategies in humans.

METHODS

Subjects

There were 15 subjects who participated in the study. These were healthy men ages 20–36 yr (25.1 ± 5.2 yr), bodyweight 72–90 kg (78.5 ± 6.6 kg), and body height 172–192 cm (179.7 ± 6.3 cm). All subjects underwent a medical examination before the experiment and, according to the Helsinki Declaration, signed informed consent for their participation in this study, which was also carried out in accordance with all tenants of the Helsinki Declaration.[23]

Equipment

The biomechanical parameters of walking at a pace of 90 steps/min were analyzed during treadmill testing using the "H/P/Cosmos Mercury 4.0" (H/P/Cosmos Sports & Medical GmbH, Nußdorf, Germany). Subjects' bodyweights were measured before each walking test using "Kistler" force plates (Kistler Group, Winterthur, Switzerland) installed under the treadmill belt. Various weight loadings on the musculoskeletal system were achieved using a vertical bodyweight support system "H/P/Cosmos-Airwalk" (H/P/Cosmos Sports & Medical GmbH) consisting of an air compressor, frame structure, and a special vest (**Fig. 1A and Fig. 1B**).

Biomechanical parameters of walking were recorded and analyzed using "VideoAnalysis Biosoft 3D" (Biosoft, Novosibirsk, Russia) in full compliance with the technique proposed by the authors for studying biomechanical parameters of locomotion before and after subjects completed a 21-d anti- and orthostatic hypokinesia condition,[22] which is presented in the procedure section below. Video recordings were performed

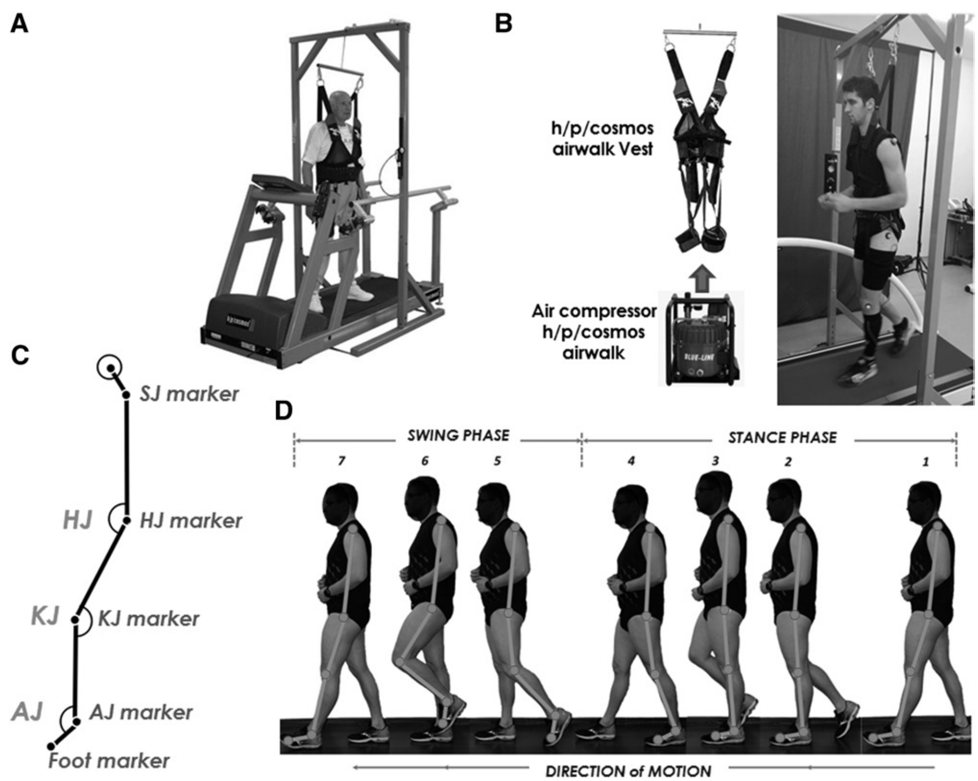

Fig. 1. A) The H/P/Cosmos Airwalk unweighting system (image from https://www.hpcosmos.com; used with permission); B) subject during locomotor test; C) four chain model connecting five main points: SJ, shoulder joint; HJ, hip joint; KJ, knee joint; AJ, ankle joint; D) the position of the lower extremities in a double step cycle from the contact of the left foot to the next contact of the left foot: 1, heel contact when placing the foot on the support (front push phase); 2, "adhesion" of the plantar surface of the foot to the support; 3, the moment of the "vertical" in the middle support phase; 4, the moment of separation of the heel from the support (the beginning of repulsion); 5, the moment of separation of the foot (completion of repulsion)—this position of the foot delimits the stance phase and swing phase; 6, the moment of passage of the maxillary foot in the plane of the supporting foot; and 7, heel contact when placing the same foot on the support (the beginning of the next step cycle).

using two Basler digital cameras, allowing video recording at a rate of 90 frames/s. A lighting system, a calibration set (Pyramid), and photographic backgrounds were also used.

Procedure

We determined a walking speed that corresponded to a pace of 90 steps/min for each individual subject using a metronome. The distance from the camera lens to the subject was 5 m. To illuminate the reflective markers, we used infrared light. For the registration of movements, light-reflecting markers were set in the shoulder joint at the level of the acromion process of the scapula and on the hip joint at the most protruding part of the greater trochanter. A marker evaluating the movement of the knee joint was fixed 2 cm above the lateral articular fissure. To register movements in the ankle joint, the lower edge of the marker coincided with the lower edge of the lateral ankle and, for the distal part of the foot, the marker was placed in the head area of the fifth metatarsal bone. The angle of the hip was measured between the longitudinal axes of the trunk and thigh from the ventral surface side of the body. The angle of the knee was measured between the longitudinal axes of the hip and shin from the dorsal side of the body. The ankle joint angle was measured between the longitudinal axes of shin and foot from the anterior surface of the shin and back surface of the foot (**Fig. 1C**). The kinematic characteristics of the locomotion were analyzed in a double step cycle (**Fig. 1D**).

Video data processing consisted of analyzing the values of the interlink angles of the hip, knee, and ankle joints. The processing of video data consisted of the analysis of the values of angles and angular velocities for the hip, knee, and ankle joints. Due to the variability of the individual kinematic characteristics of the walking during data processing, we first analyzed 15–20 consecutive double steps. Then, the duration of each double step was taken as 100%, thus translating from absolute values of time to relative. At each relative time point, we obtained averaged values of all target variables. The subject's body was considered to be a flat four-chain model connecting five main points: the acromion of the scapula (acromion), the large trochanter (trochanter major), the lateral epicondyle (epicondylis lateralis), the lateral malleolus (malleolus lateralis), and the third phalanx of the left toe on the sagittal plane of the body (Fig. 1C).

Each subject performed three walking tasks with different gravitational musculoskeletal system loads. These included:

- Walking under 100% bodyweight corresponding to terrestrial conditions;
- Walking under 38% bodyweight, corresponding to gravitational conditions on the surface of Mars; and
- Walking under 17% bodyweight, corresponding to gravitational conditions on the lunar surface.

Certain segments of the subject's body were considered undeformable; joints were replaced by hinges without friction (ideal). The mass-inertia characteristics of each subject were unchanged during the test.

To evaluate the locomotion strategy in modeled gravitational conditions, we constructed various phase trajectories for the hip, knee, and ankle joints. All phase trajectories were constructed in such a way the angles in the joint were plotted on the abscissa axis, while the angular velocity was plotted on the ordinate axis. The phase trajectory areas were estimated without bodyweight support (walking under 100% bodyweight), walking under 38% bodyweight, and walking under 17% bodyweight.

In addition to the analysis of joint phase trajectories, we analyzed energy costs; that is, joint angular velocities. A similar study was carried out under the conditions of a 7-d "dry" immersion at the Institute of Biomedical Problems (Moscow, Russia).[17] In this study, locomotion patterns were comparatively analyzed before and after dry immersion exposure. For this study, we used previously published energy cost calculations of energy costs.[25] The kinetic energy of rotational motion, which was associated with joint flexion and extension, was calculated as:

$$E_i^{\pm} = \frac{1}{2}\sum_k \left(J_i + mr^2\right) * \left(\varpi_{i,k}^{\pm}\right)^2 \qquad \text{Eq. 1}$$

Where J_i equals inertia moment in the *i*-link segment, *m* equals link weight, *r* is the distance from the mass link center to the point of suspension, and $k = 1$ the weight load of 100%, $k = 2$ the weight load of 38%, and $k = 3$ the weight load of 17%. For defining the kinetic energy of flexion and extension in *i*-link, angular velocities were divided into positive [$\varpi_{i,k}^{+}$ (extension)] and negative [$\varpi_{i,k}^{-}$ (flexion)] ones.

The ratio of the kinetic energy of extension (E_i^+) to the kinetic energy of flexion (E_i^-) for the *i*-joint was calculated using the formula:

$$R_i^{\pm}(k) = 100\% * E_i^{+}(k) / E_i^{-}(k) \qquad \text{Eq. 2}$$

To better understand locomotion strategies, we analyzed and compared energy changes [$D_i^{\pm}(k)$] in flexion/extension motions in walking when bodyweight was 100%, 38%, and 17%. The obtained results were calculated using Eq. 3:

$$D_i^{\pm}(k = 2,3) = 100\% * E_i^{\pm}(k = 2,3) / E_i^{\pm}(k = 1) \qquad \text{Eq. 3}$$

Calculation of phase trajectory area is one method for describing motion strategy. We assessed the area of phase trajectories [$S_i^{\pm}(k)$] for all joints and all levels of bodyweight support. We also found relative values as a percentage value [$F_i^{\pm}(k)$]. In all cases, we determined the ratio of phase trajectories to those but without bodyweight support (Eq. 4).

$$F_i^{\pm}(k = 2,3) = 100\% * S_i^{\pm}(k = 2,3) / S_i^{\pm}(k = 1) \qquad \text{Eq. 4}$$

Statistical Analysis

The Statistica 8 (Statsoft, Tulsa, OK, USA) software package was used for all statistical calculations. The reliability of differences was determined using the nonparametric Wilcoxon test. Results which achieved a level of significance of $P < 0.05$ were considered reliably significant. The reliability of the obtained results is found in notes to the corresponding tables and figures.

RESULTS

Fig. 2 shows typical phase trajectories which characterize hip, knee, and ankle joint movements. Similar phase trajectories were met in 11 out of 15 subjects (group No. 1). The remaining four subjects (group No. 2) showed a marked displacement of phase trajectories in angular coordinates during maximal reduction of weight loading (up to 17% bodyweight support).

All phase trajectories, except those of the ankle joint, exhibited progressively reduced phase trajectory areas alongside simultaneous reductions in weight loading. To compensate for these changes, ankle joint movements were redistributed. Consequently, phase trajectories for the ankle joint increased in areas of decreased loading.

One important feature of knee joint kinematics is the "amortization" phase (i.e., a slight bending and angle reduction in the joint at the beginning of the double step cycle which was 10–15% of double step time. Here, bodyweight is completely transferred to the supporting leg (time interval from the initial contact to the moment of the "vertical" phase trajectory of the knee joint). Thus, while walking with 100% of bodyweight, all subjects exhibited a small loop beginning at the front push phase to the "vertical" phase in the middle support phase. The loop became smaller as bodyweight support was reduced to 38%. When bodyweight support decreased to 17%, the phase trajectory loop completely disappeared. Such changes in the articular kinematics indicate that, while walking with 17% bodyweight support, there is no need for additional knee joint motion during the "amortization" phase under reduced weight loading.

Fig. 3 shows phase trajectories for four subjects. We observed phase trajectory displacement as joint angles decreased under decreased bodyweight loading, up to 17%. Such locomotion (walking) strategy can be explained by the fact that these subjects completely relied on the bodyweight support system and assumed a semibent posture. The most striking difference was seen for phase trajectories of the hip and ankle joints. When walking with 17% bodyweight support, the hip and knee joint phase trajectories varied approximately equally. So, walking with 38% bodyweight support, the relative changes in the area of the phase trajectory for the hip and knee joint were 52.70 ± 8.88 and 54.44 ± 8.47, respectively. A similar match was observed during walking with bodyweight support up to 17%. The

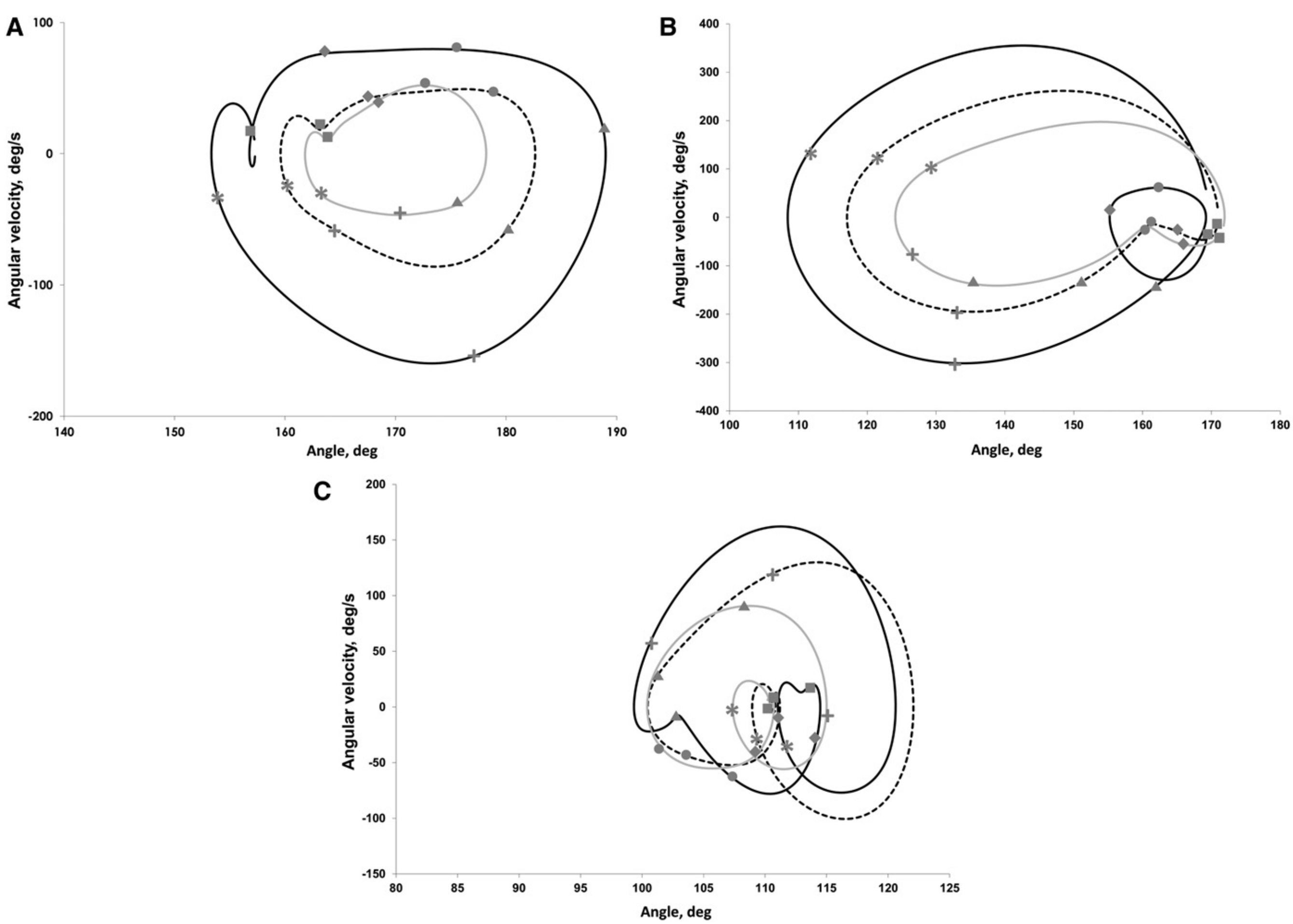

Fig. 2. Normalized phase trajectories of the A) hip, B) knee, and C) ankle joints during walking under various weight musculoskeletal system loads. Black line: walking with 100% bodyweight support; dotted line: 38%; gray line: 17%. Square: initial contact; diamond: foot flat (opposite toe off); circle: 'vertical' moment in the middle support phase; triangle: heel off (beginning of push-back phase); plus symbol: toe off (repulsion is almost completed; push-back phase); asterisk: feet adjacent. The cycle then starts over.

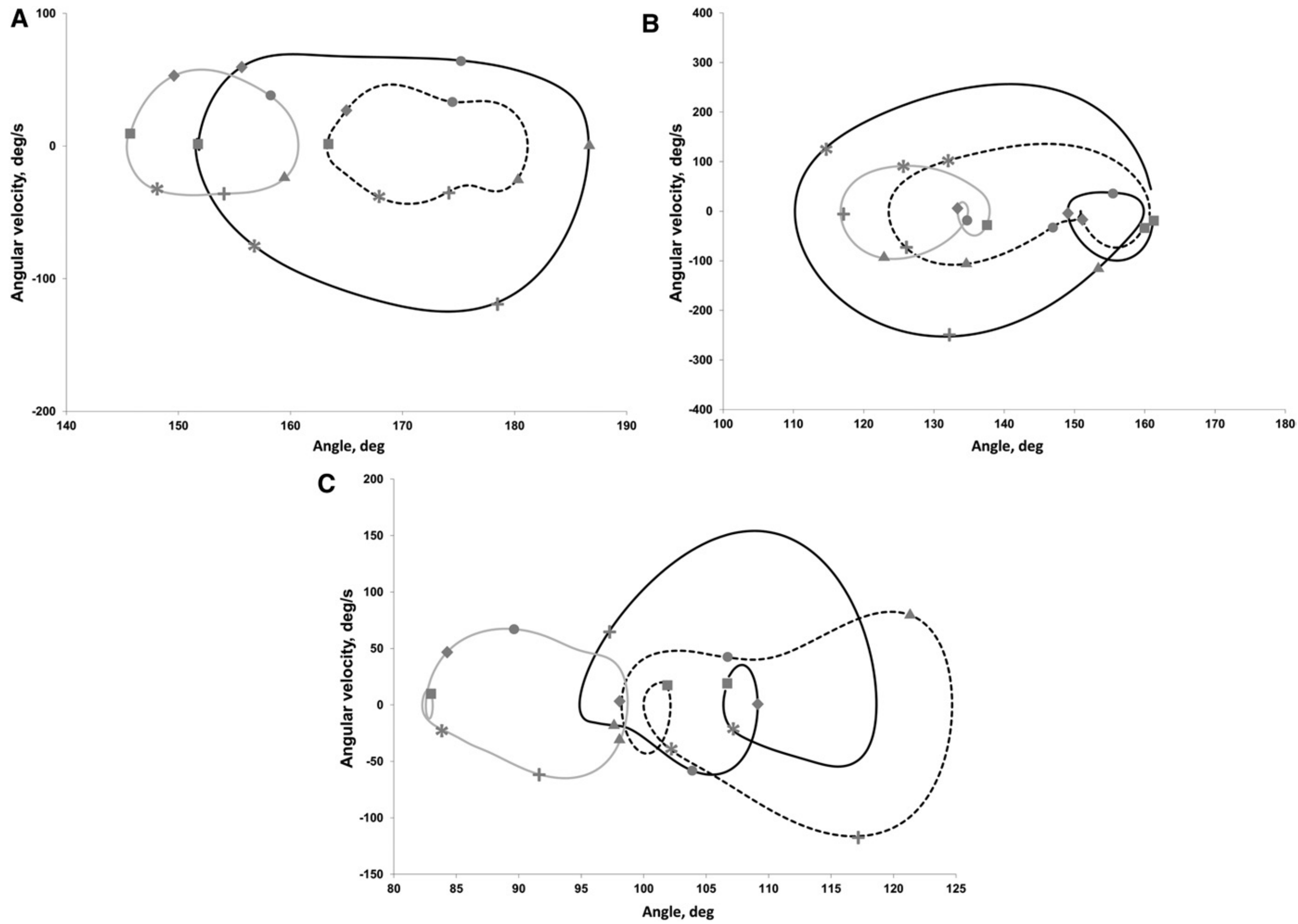

Fig. 3. Normalized phase trajectories of the A) hip, B) knee, and C) ankle joints during walking under various weight loadings on the musculoskeletal system. Black line: walking in 100% bodyweight support; dotted line: 38%; grey line: 17%. Square: initial contact; diamond: foot flat (opposite toe off); circle: 'vertical' moment in the middle support phase; triangle: heel off (beginning of push-back phase); plus symbol: toe off (repulsion is almost completed; push-back phase); asterisk: feet adjacent. The cycle then starts over.

change in the hip joint was 33.18 ± 7.27 and the knee joint was 35.39 ± 8.85 (**Table I**). Thus, one movement strategy appeared to involve a relative change in the area of the phase trajectory, both for the hip and the knee joints. The movement strategies involving these two joints appear closely related.

Table II presents group averages for the ratio $R_i^{+/-}(k)$ of flexion and extension kinetic energy for different levels of weight loading -*k* (loading level for the locomotor system is indicated the same way as in Eq. 1). In terrestrial conditions (walking under 100% bodyweight), flexion kinetic energy prevailed over extension kinetic energy by 90%, with a small variation equal to 22%. If weight loading decreased up to 17% (lunar conditions), the difference between flexion and extension kinetic energies diminished and eventually reached only 9%. However, in the ankle joint, we observed the opposite phenomenon. The ratio of flexion energy and extension energy equalized under lower loading conditions. Thus, 38% bodyweight loading was sufficient for approximation of flexion and extension energy values.

Parameter $D_i^{\pm}(k)$ shows how much energy of normal loading constitutes an extension or flexion energy under various loading levels (**Table III**). Kinetic energy for hip and knee joints in bodyweight support, up to 38% of bodyweight, was about 50–60% of that without bodyweight support (100% bodyweight). In 17% bodyweight support, these energies values did not exceed 50% and averaged 35% of the corresponding kinetic energy without bodyweight support. However, it should be

Table I. Average Data for Group No. 1 ($N = 11$) of the Ratio of Phase Trajectory Area in Bodyweight Support to the Same Parameters During Normal Walking [$F_i^{\pm}(k)$].

LOAD, % BODYWEIGHT	HIP JOINT	KNEE JOINT	ANKLE JOINT
38%	52.70 ± 8.88	54.44 ± 8.47	130.08 ± 32.04
17%	33.18 ± 7.27*	35.39 ± 8.85*	90.32 ± 20.01

* Significant differences compared with walking at 100% of bodyweight ($P < 0.05$).

Table II. Average Data for Group No. 1 ($N = 11$) of the Ratio of Extension Kinetic Energy to Flexion Kinetic Energy for Different Joints [$R_i^{+/-}(k)$].

RATIO, %	HIP JOINT	KNEE JOINT	ANKLE JOINT
k = 1; 100% bodyweight	190 ± 22	89 ± 7	47 ± 11
k = 2; 38% bodyweight	143 ± 28	74 ± 7	97 ± 29
k = 3; 17% bodyweight	109 ± 29	77 ± 15	99 ± 23

Table III. Average Data for Group No. 1 ($N = 11$) of the Ratio of Extension and Flexion Energy Under Decreased Weight Loading (38% and 17% Bodyweight) to the Same Parameters in Normal Walking [$D_i^{\pm}(k)$].

LOAD, % BODYWEIGHT	ENERGY	HIP JOINT	KNEE JOINT	ANKLE JOINT
38%	flexion	48.07 ± 22.93	51.01 ± 25.40	206.82 ± 137.64
	extension	64.13 ± 31.02	60.22 ± 24.51	99.28 ± 59.56
17%	flexion	28.01 ± 13.70	34.36 ± 17.41	147.10 ± 72.05
	extension	48.88 ± 24.72	39.82 ± 18.26	69.82 ± 30.21

noted that the most significant reduction in weight loading affected hip joint flexors. Thus, maximal bodyweight support, up to 17%, led to a decrease in flexion energy kinetics in the hip joint by about 3.5 times. In contrast, flexion kinetic energy in the ankle joint increased by about 1.5–2 times. This indicated that there was a redistribution of flexion and extension energies at various load levels within the musculoskeletal system.

DISCUSSION

According to data obtained, all subjects were divided into two unequal groups. The first group (group No. 1) consisted of 11 subjects with nearly identical phase trajectories in the lower extremity joints. This group had a locomotion strategy wherein reduced loading led to reductions in phase trajectory area for both the knee and hip joints. This group also exhibited fewer variations in angles and angular velocities in these joints, while the phase trajectory area in these joints was reduced proportionally to musculoskeletal system loading. However, the variability of ankle joint kinematics increased as weight loading decreased. These changes in ankle joint kinematics may compensate for the number of movements in the hip and knee joints under bodyweight support since it allows preservation of the speed and pace of walking. This strategy is likely to be dominant and typical for most subjects.

Out of 15 subjects, 4 were included in the second group (group No. 2). When the weight load on the musculoskeletal system was reduced to 17% of bodyweight, some subjects fully relied on the bodyweight support system. In these cases, we observed reduced interlink angles in the hip, knee, and ankle joints. Consequently, the subject's posture became "slightly bending." Such posture is not stable and not particularly suited to effective walking in these conditions because of restricted movement. Consequently, these conditions violate correct locomotion strategies. Similar changes are observed in phase trajectories as well. So, in cases of maximum bodyweight support (reduction of weight loading up to 17% of bodyweight), phase trajectories shifted toward smaller joint angles.

Our results also revealed a decreased ratio between extension kinetic energy and flexion kinetic energy in the knee joint of all subjects. With no bodyweight support (100% bodyweight), this ratio was 190% (i.e., extension kinetic energy was about twice as large as flexion kinetic energy). When musculoskeletal system loading was reduced up to 38% of bodyweight, this figure became 143%. Finally, in conditions involving maximum bodyweight support and 17% body loading, this ratio was practically equal. Reverse dynamics were seen in the ankle joint. We did not find significant differences in the ratio of flexion and extension kinetic energy in the knee joint under bodyweight support.

Decreased joint kinetic energy clearly depended on the degree to which weight loading was decreased. In contrast to these changes, flexion kinetic costs were increased in the ankle joint. This, once again, supports our observed strategy: subjects used less energy in the knee and hip joints and more flexion kinetic costs in the ankle joint to maintain the chosen walking tempo and speed.

These results have implications for training of astronauts for future interplanetary expeditions. Understanding the distinctive features of normal walking and walking in conditions of reduced gravitational loads, we can make targeted changes to methods used to train crew. It should be borne in mind that the most pronounced changes occur in the movements of the ankle joint. Therefore, difficulties should be expected when walking in a spacesuit with hard boots in conditions of reduced gravity (loss of balance, a change in the usual pace of walking, etc.). We propose to take into account the need for high mobility in the ankle joint when designing new types of spacesuit. In addition, based on the work done, a cosmonaut training scheme can be proposed in preparation for interplanetary missions such as training of locomotion on a treadmill in the conditions of hanging up to the required level of gravitational load and locomotion training in hard/mobile boots with recommended walking intensity. To create the correct stereotype of walking, it is necessary to perform these exercises in the type of shoe that will be applied in the design and manufacture of the spacesuit. Additionally, the methods used for musculoskeletal system assessments that were proposed in our work can be used in clinical practice to evaluate the effectiveness of rehabilitation measures in a patient's musculoskeletal system disorders (for example, sports injuries, cerebral palsy, and stroke rehabilitation).

In conclusion:

1. Conditions of reduced musculoskeletal system loading change locomotion strategy, as confirmed by changes in leg joint phase trajectory areas.
2. The results obtained demonstrate different locomotor strategies are used under various gravitational conditions.
3. The first strategy was dominant in all subjects. We observed fewer angle variations and angular velocities in the hip and knee joints under reduced weight loading. At the same time, we observed increased angular kinematic variability in the ankle joint. This strategy ensures posture stability while walking.

4. The second locomotion strategy was characterized as a slightly bent posture during walking. This adaptation was less stable and effective in response to changes in musculoskeletal system weight loading. However, it should be noted that such a strategy may be characteristic of exactly four subjects. And this was the first attempt to characterize the strategy of human locomotion and unloading bodyweight. To obtain more statistically reliable results, it is necessary to carry out studies with a large number of subjects of different age groups.
5. When musculoskeletal system weight loading changes, the ratio between extension and flexion kinetic energies also changes.
6. In the hip joint, while walking under 100% bodyweight, flexion energy prevails. With bodyweight loading reduced up to 17%, the ratio between extension and flexion energies comes into balance.

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