‘APPLICATION OF BIOMECHANICAL METHODS TO THE STUDY OF HUMAN LOCOMOTION AND ATHLETIC ACTIVITIES’

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The major characteristic of human locomotion is that only in exceptional cases does it proceed along a straight line. The alterations in direction of movement and the alterations in speed of movement correspond to accelerations to which all body parts are subject. To produce accelerations externally imposed forces are required and, in fact, a study of displacements without a knowledge of the forces or a study of forces without a knowledge of the displacements gives very little further knowledge about the mechanics of the system of progression. It is convenient in discussing body mechanics to think in terms of the centre of gravity of the body which is that point at which the whole mass of the body may be considered to be concentrated in respect of its response to external applied force.

During walking on a level surface the centre of gravity of the human, if he is normal, rises and falls twice per cycle through a distance of approximately two inches and sways from side to side once per walking cycle, again through an amplitude of approximately two inches. Thus the centre of gravity is subject to accelerations in the vertical direction and accelerations in the medio-lateral direction and the patterns of variation of ground to foot force must correspond to this.

A more subtle phenomenon is the variation in the forward speed of progression. For a patient walking in a straight line forwards at an average speed measured over a considerable distance of, say, S miles per hour, he fluctuates in forward speed from a maximum of 115% of S to a minimum of 85% of S during each cycle of locomotion. Thus one would expect as well as vertical and medio-lateral forces between the ground and the foot that there would also be anterior/posterior component of ground to foot force. One might expect that this would be necessary anyway in order to maintain forward progression but, on a level surface, and in the absence of a contrary wind there is very little resistance to forward movement and the front to back ground foot force is almost exclusively concerned with producing the variations in forward velocity.

The walking cycle can best be described in terms of the recognisable events during it. Starting from, say, heel strike of the left foot the next event for the left foot will be when it leaves the ground at toe off, it will then progress through the swing phase until the next heel strike of the left foot. This cycle from heel strike of left foot to next heel strike of the same foot is taken to be 100% of full cycle. For a symmetrical gait, heel strike of the right foot would occur at 50% of this cycle and, to

talk in round numbers, it is found that the toe-off of the left foot would occur at approximately 60% on the cycle, that is, the swing phase occupies 40% of the cycle. For the right foot therefore, its toe-off would be 40% before heel strike, that is 40% before 50% of the cycle being discussed - this is at the 10% mark measured relative to the events of the left foot. In fact, these numbers that I quote here are by no means constant and vary particularly with the speed of locomotion. The slower the walking speed the longer the support phase and the shorter the swing phase. It will be noticed that there are two phases of about 10% each during which the body is supported by both feet and, as walking speed increases, this overlap period of double support diminishes progressively and indeed, in race walking, one of the difficult decisions to be made by the judges is whether walking is in fact proceeding and there exists this period of double support.

In the measurement of ground to foot force the most usually adopted procedure is to construct a dynamometer, or in fact several dynamometers, and install them in recesses in the floor surface so that they are flush with the floor surface, the intention being that successive foot falls of the test subject should land on separate dynamometers. Since many people when walking have a very narrow lateral tracking distance between feet, this means that it is impossible to arrange dynamometers which will cater one for each foot only, although such instruments have been constructed and are in use at some centres. The primary disadvantage is that the constraint on the test subject to walk with his legs slightly apart and this is, of course most abnormal for the majority of walking subjects.

The alternative then is to make each dynamometer sufficiently small to accommodate one foot only and yet sufficiently large to make sure that the full period of foot contact takes place while the foot is on the dynamometer. Obviously one cannot train all the test subjects so that the foot lands in appropriate and correct places. Most centres use two plates arranged so that successive foot falls of left and right foot are measured. Another important factor here is that the human is very sensitive to changes in the supporting surface and the dynamometer therefore must be sufficiently rigid not to deflect more than one thousandth of an inch in any direction when the foot lands on it and, of course, to be sufficiently robust to withstand the forces exerted on it. Another difficulty of any measuring system is the adequacy of its response to dynamic signals. This is generally best judged by the speed at which the

instrument vibrates by itself if set into vibration. It is generally conceded that the ideal minimum frequency of vibration should be five hundred cycles per second, although very few of the force plates currently available can equal this ideal performance.

The record obtained for the vertical component of ground to foot force has a characteristic double peaked shape extending from zero at heel strike up to of course zero again at toe-off. The two maxima correspond very closely to the times when the contra-lateral foot is itself leaving the ground and regaining it. The values of these maxima are approximately 35 to 40% greater than body weight and between them is a trough of some 60 to 70% of body weight corresponding to the period during which the body is accelerating towards the ground having passed its highest point. The front to back component of force is generally initially directed on the foot in a direction opposite to the progression direction and it remains of this sense up to approximately the middle of stance, when it changes over to become forwards directed on foot. In the overlap during double support there is simultaneously forwards force on the rear foot, together with a backward force on the forward foot, so that these two components are working against each other for part of the cycle. The sideways force by the ground on the foot is much smaller than the others in magnitude being at maximum approximately 7% of body weight and is for most of the cycle directed medially.

From the records of displacement of the limb segments giving instantaneous position at various stages in the locomotion cycle and the known values of the forces, one can infer the position of the resultant ground force relative to areas of interest, particularly the joints, and one can therefore infer the forces transmitted by ligamentous structures and by muscle. Further, from viewing successive instants in time, one can recognise the mode of action of the muscles, whether they are loaded isometrically or are exerting force and contracting or exerting force and extending, that is absorbing energy. This interpretation however is open to question in the assumptions it makes in respect of the muscles which are contributing to any load action at any time, in that most joint movements correspond to the simultaneous action of several muscles whose resultant effect is very much the same, and the allocation of the share of load to individual muscles is generally not possible. Certainly myo electric signals can be measured and the presence or absence of stimulation signal on the recorder can be used as an indication whether a muscle is in fact participating in a particular movement. As far as the gross muscles of the leg are concerned, in most normal activity all synergistic muscles are stimulated to some degree at the same time and the art of electro-myography has not yet progressed far enough to be able to allocate a share of load to the individual muscles corresponding to the

intensity of the EMG signal. The interpretation of the EMG is of course particularly complicated since a given level of signal can correspond to quite different forces depending on the velocity of contraction of the muscle, its instantaneous length relative to a standard position and, also, on whether it is at that instant lengthening or shortening, as well as on other metabolic factors such as fatigue and so on. Even the interpretation of periods of muscular activity in the respect of EMG records is fraught with a difficulty which is not always recognised in that there is a time delay between the onset of EMG activity and the initiation of force and, correspondingly, a similar delay between the termination of EMG activity and the diminution of muscle force to zero.

There is another instrument which appears to have possibilities in respect of the recording of locomotion characteristics and that is the accelerometer, which of course measures the acceleration, that is the rate of change of velocity, along the sensitive axis of the instrument. These instruments are now commercially available of suitable sensitivity and suitably small bulk and mass. As with all improvements of this type in measurement, there are disadvantages however, in that the accelerometer gives only the signal corresponding to acceleration in the direction in which it is pointing, and since all limb segments rotate about some axis during activity, it is very difficult to get exactly the acceleration one wants. Furthermore, the accelerometer gives another signal corresponding to the inclination of its sensitive axis to the vertical. This is due to the effect of gravity and this cannot be cancelled out. Probably the major disadvantage of accelerometers is the fact that they have to be secured to the body. One would probably hope to be able to measure the accelerations of the skeletal system but the trouble is that there is always intervening tissue between the skeleton and the skin surface and it is particularly difficult to immobilise an accelerometer relative to the skeletal structure without applying some constraint to the user.

As was mentioned in respect of force plates it is important to consider the dynamic behaviour of accelerometers in that they are an instrument with mechanical parts which can vibrate and therefore their natural frequency of vibration will control to a large extent their usefulness in dynamic recording. Generally, the more sensitive an instrument, the lower will be its natural frequency, but one would want a natural frequency of vibration of an accelerometer to be at least 500 cycles per second, even though you are measuring locomotion, which has a basic speed of approximately one cycle per second. The other difficulty with accelerometers is that each instrument will measure only the quantity for the point to which it is attached, and there are so many segments of the body and directions in which they can move that to measure all by accelerometers would mean attaching an impossible load to the performer.

To summarise my philosophy of this type of research, I must say that I would think it essential to record spatial configuration of the trunk and limbs, the forces transmitted externally to the body and the phasic activity of the muscles as measured by EMG. In the

absence of information on all these quantities I feel that only an incomplete analysis could be made, and, in my opinion, the acquisition of information less detailed that this can lead only too easily to erroneous conclusions.

CHANGES IN HEART MOVEMENTS DURING CONTROLLED PHYSICAL ACTIVITY

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Any significant physical activities immediately accompanied by an increase in both heart rate and stroke - volume. The more severe the exercise the greater the changes are observed but the fitter the individual the more difficult it is to take consistent results at low levels of loading and the more difficult it is to extract quantitative information about the functioning of the heart.

This paper is concerned with a new method of studying heart action during exercise which involves the detection and recording of changes in the movement of the heart particularly at the apex which is proving to be a very useful parameter.

The principle of the method lies in measuring changes in magnetic inductance using a probe coil which is part of an oscillator and is placed over the apex. Tissue displacement under the probe, which need not be in contact with the skin, are converted to electrical signals which can be transmitted to any standard recorder. The displacements observed are complex in origin and detailed features of the record can be associated with other known cardiological events.

RESULTS OBSERVED AT REST

Fig. 1 shows a typical apex movement as recorded above the torso of a normal subject in basic metabolic rate (approx. 90Kcal/hr.). The 'a' wave represents the

Fig. 1

apex recoil during atrial contraction. It is said that the ascending part of the 'a' wave represents the contraction of the right atrium and the descending part of the 'a' wave stands for the left atrial contraction. The isovolumetric contraction (some call it the pre-systolic wave) is shown on the heart movement as the ascending part of the E wave. The E point itself marks the opening of the aortic valve and therefore the beginning of the ejection phase of the heart. In this period the heart ejects the blood in the aorta. The O point represents the opening AV valves i.e. the beginning of the diastole which in turn ends at the onset of the 'a' wave. Two

Fig. 2