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Biomechanical analysis of standard and wheelchair Basketball set shooting

University — Université

of ALBERTA

Degree for which thesis was presented — Grade pour lequel cette thèse fut présentée

MSc

Year this degree conferred — Année d'obtention de ce grade	Name of Supervisor — Nom du directeur de thèse
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NL-91 (4/77)

THE UNIVERSITY OF ALBERTA

BIOMECHANICAL ANALYSIS OF STAND-UP AND WHEELCHAIR BASKETBALL SET SHOOTING

by

YUVAL HIGGER

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE

OF MASTER OF SCIENCE

DEPARTMENT OF PHYSICAL EDUCATION AND SPORT STUDIES

EDMONTON, ALBERTA

FALL 1984

THE UNIVERSITY OF ALBERTA

RELEASE FORM

NAME OF AUTHOR YUVAL HIGGER

TITLE OF THESIS BIOMECHANICAL ANALYSIS OF STAND-UP AND WHEELCHAIR BASKETBALL SET SHOOTING

DEGREE FOR WHICH THESIS WAS PRESENTED MASTER OF SCIENCE

YEAR THIS DEGREE GRANTED FALL 1984

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ABSTRACT

The purpose of this study was to examine selected kinematic and kinetic factors in the performance of the one hand basketball set shot as performed by stand-up basketball players and wheelchair basketball players of different levels of physical disability. Kinematic information was obtained from the anlysis of data films whereas resultant muscle torques were computed using principles of rigid body dynamics. For this purpose individuals' body segment parameters were estimated. It was found in this study, that wheelchair basketball players projected the ball with a greater speed of release and a higher angle of release than did stand-up basketball players. Nevertheless, balls projected by wheelchair basketball players approached the rim with a slightly smaller angle of approach. It was also found that wheelchair basketball players generated greater muscle torques in order to propel the ball towards the rim. However, the increased torques were not proportionately distributed. Shoulder flexion and shoulder extension torques of wheelchair basketball players were relatively greater than their elbow extension and wrist flexion torques. Class II wheelchair basketball players and class III wheelchair basketball players projected the ball employing almost identical trajectories. To do so, class II and class III wheelchair basketball players generated similar shoulder

flexion and shoulder extension torques, but class II wheelchair basketball players generated smaller elbow extension and wrist flexion torques. Considering the differences in body segment parameters between class II and class III wheelchair basketball players in this study, it was speculated that when compared to class III wheelchair basketball players, class II wheelchair basketball players generated relatively greater shoulder flexion and shoulder extension torques than elbow extension and wrist flexion torques.

ACKNOWLEDGEMENTS

The author wishes to express sincere appreciation to:

Dr. E. W. Bedingfield who taught, advised, motivated and encouraged - THANKS WENDY,

Pierre Gervais who introduced me to the program and shared with me of his extensive knowledge

Dr. R. Steadward and Dr. G. Smith, other members of the committee

A fellow graduate student Gord Marchiori, who provided me with so much help

Special thanks are extended to our family Alex and Miri Robson who made our stay in Edmonton "A HOME" away from home. THANKS Alex, THANKS Miri

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I. INTRODUCTION

Basketball is one of the most popular games in the world. Stand-up basketball (which is the original game and therefore will be defined as "basketball") is the most popular winter game throughout the United States, and is secondary only to soccer in Europe and South America. The game is played as a recreational activity as well as in a highly competitive manner by both males and females. Competitively, basketball is played by high school, college, amateur and professional teams throughout the world (Hobson, 1955).

Similarly, wheelchair basketball is very popular among the physically disabled. It has emerged as the only wheelchair game that is played worldwide, with an increasing number of individuals participating. The game was first played at the V.A. hospitals in New England and in California in 1946, where it was initially used as a means of recreation and rehabilitaton. With the formation of the National Wheelchair Basketball Association (N.W.B.A) in 1949, wheelchair basketball has become a competitive sport with growing participation. By the end of 1981, the N.W.B.A had a membership of 136 male teams playing in 25 conferences throughout the United States. Eight women's teams participate in a separate division (Owen, 1982).

The game of wheelchair basketball is a modification of stand-up basketball. Shavor (1981) suggested that the wheelchair game is more like than unlike "normal" basketball. The rules are the same, apart from a few changes to accommodate the use of the wheelchair and the various degrees of disability. Also, the desire to encourage part[illegible]ation by more individuals with different levels of disability has led to the development of a player classification system.

The basic playing skills of shooting, passing and dribbling are fundamentally the same in both versions of the game (Skillen, 1983). However, in the wheelchair game the original skills are slightly m[illegible]ified, as the players are positioned lower, and the propulsive forces are derived mainly from the arms and upper body. Due to the lower body disability, very little if any of this force will come from the legs. The question arises as to how skills should be modified in an attempt to optimize performance of physically disabled athletes. This question warrants scientific investigation.

Biomechanics, as one of the many sport-related sciences, has made a significant contribution to the field of physical education and sports. Scientific tools have been developed and widely used in an attempt to assist athletes and coaches striving towards better performances. Unfortunately, despite the definite need, biomechanics has yet to make its contribution to the continually growing area of sports and physical education for the physically disabled. To date, few studies have been published in which wheelchair events are biomechanically analyzed. The need for such studies is obvious.

The author attempted to biomechanically analyze a basketball-related skill as performed by wheelchair players, and to compare it to the equivalent skill as performed by physically able, stand-up basketball players. Shooting was chosen since it is a key skill in the game of basketball. Mitchelson (1978) described it as the prime skill in the game. He suggested that the remainder of offensive and defensive skills exist only to assist or prevent scoring opportunities. Also, it is a highly technical skill, as athletes must devote a great deal of their time developing consistent shooting technique. From the many typical shots, the one-hand set-shot from the free throw line was chosen to be studied for the following reasons .

1. The free throw is an important aspect of the game. Hobson (1955) claimed that championships are won at the free throw line. Hobson explained his argument stating that nearly one-fourth of all games played are decided by four points or less. In each of these games a perfect free throw record by the losing team would have made them the winners. Even though it seems that Hobson has slightly over emphasized the importance of the skill, free throws are important, and therefore deserve biomechanical analysis.

2. One hand shooting is the most common shooting technique in basketball, adopted by almost all competitive basketball players. The one hand set shot is the basic shooting technique and its principles are consistent for all one hand shots (Mitchelson, 1978).
3. The free throw line is located at a horizontal distance of 4.37 m from the centre of the rim. Consequently, the free throw is a mid-range shot. In the game situation, many shots will be taken from such a range in both stand-up and wheelchair basketball.
4. The free throw is always performed in a static position, is never defended against, and is always shot from the same position on the court.

In spite of the similarity that exists between the two versions of the skill, highly skilled stand-up basketball players perform better than wheelchair players from the free throw line. Numerically, during a five year span ,(1950-55), high school players throughout the United States achieved a 65% success rate from the free throw line, N.C.A.A. players scored in 63% of their attempts, and professional athletes shot 72% (Hobson, 1955). Today, players achieve even slightly better results. On the other hand, in wheelchair basketball, scoring averages from the foul line range between 45 and 55 percent (Owen, 1982). Data from the final rounds of the 1981 N.W.B.A Championship showed that the best four teams achieved a 40% rate of success.

Several reasons could account for this discrepancy in the performance of fundamentally the same skill. For example, data might simply indicate that stand-up players are more skilled, better trained, or better mentally prepared. However, it was suspected that some mechanical factors make the execution of a free throw in wheelchair basketball more difficult. These mechanical disadvantages are probably related to either one or both of the following factors :

1. The seated position of the athlete.
2. The physical disability of the individual.

The mechanical disadvantage that results from the lowered seated position equally affects all wheelchair basketball players. However, the severity of the physical disability differs from person to person, and thus, athletes are subjected to varying limitations which unequally affect their performance. Consequently, a system of classification was developed to account for different degrees of disability (Appendix A).

A. Statement of the Problem

The purpose of this study was to examine selected kinematic and kinetic factors in the performance of the one hand set shot as performed by wheelchair basketball players of different disability levels, and to compare the performance of wheelchair athletes to the performance of the one hand set shot executed by physically able stand-up basketball players. Analysis and comparison concentrated on:

1. The trajectory of the ball.
2. The kinetics of the shooting upper limb.

The investigator hypothesized that:

1. Wheelchair basketball players would project the ball at a greater angle of release and with a greater speed of release, and that the ball would approach the rim at a smaller angle of approach.
2. Wheelchair basketball players would generate greater flexion torques at the wrist joint, greater elbow extension torques and greater shoulder flexion torques than would stand up basketball players.
3. Both class II and class III wheelchair athletes would propel the ball with a similar speed and angle of release.
4. Class II basketball players would be required to generate larger torques at the wrist, elbow and shoulder joints than would class III athletes.

B. Limitations

The following are the limitations of this study:

1. Motion occurs in three planes. This study was a two-dimensional analysis of motion, therefore movements out of the plane parallel to the film were not measured.

2. The accuracy of determining segment length was limited to the accuracy of the Humanscale Anatomical Data (Diffrient, ~~1979~~), and ~~to the~~ researcher's ability to locate the proximal and distal end points of the body segments. Other body segment parameters (segment weight, center of mass and radius of gyration) were estimated based on cadaver studies (Dempster 1955 as cited by Winter 1979). The accuracy of the obtained parameters was limited to the accuracy of Dempster's coefficients.

3. The errors inherent in cinematographical data acquisition and analysis were unavoidable even though precautions were taken to minimize perspective errors, film graininess and distortions through the optical elements of the recording and/or projecting procedures (Appendix D).

4. The accuracy of the kinetic analysis was limited to the validity of the following assumptions, made for the derivation of the equations of motion:

    a. Each segment had a fixed mass located as a point mass at its center of gravity.

    b. The location of the segment center of mass remained fixed during the movement.

    c. The mass moment of inertia of each segment about its center of mass was constant during the movement.

    d. The joints were hinge joints.

e. The discrepencies introduced by two joint muscles were negligible.

C. Delimitations

This study was delimited in the following ways:

1. Subjects in this study were wheelchair basketball players classified as Class II (n=10), and class III (n=10) wheelchair basketball players, and stand-up basketball players (10 subjects).
2. The cinematographical analysis was restricted to a two dimensional analysis of motion in the sagittal plane.
3. A sampling frequency of 150 frames/second was utilized.
4. Right side views of all performances were filmed. Consequently, only right handed performances were analyzed.
5. One shot per subject was filmed. Of the filmed shots, only successful and close shots that hit the rim were analyzed.

D. Definition of Terms

Class II wheelchair athletes are physically disabled athletes who are classified as class II according to the players classification system of the N.W.B.A This classification system is based upon the level of lesion and the degree of disability of wheelchair athletes (Appendix A).

Class III wheelchair athletes are physically disabled athletes who are classified as class three according to the N.W.B.A classification system (Appendix A).

Ball release is the first instance of time when contact between ball and finger tips is lost.

Angle of release (θr) is the angle formed by the tangent to the ball's center of mass pathway and the horizontal at release.

Angle of approach (θa) is the angle formed by the tangent to the ball's CM pathway and the horizontal at the moment the lowest part of the ball reaches rim height.

Speed of release (Vr) is the instantaneous speed of the ball's center of mass at release.

Angular velocity (ω) is the rate of change of angular displacement about a joint.

Angular acceleration (α) is the time rate of change of angular velocity.

Propulsive phase is the time interval during which the ball is accelerated prior to release.

Predicted body weight is the body weight of a physically disabled athlete, plus the weight of the missing or reduced segments.

Torque is the product of the force and its moment arm. A torque tending to generate motion in the counterclockwise direction was defined to be positive torque, whereas a negative torque was a torque that acted in the clockwise direction.

Segmental angles (ϕ_1, ϕ_2, ϕ_3, ϕ_4) are the angles formed by the long axis of the trunk, arm, forearm and hand respectively, with the right horizontal (Figure 1).

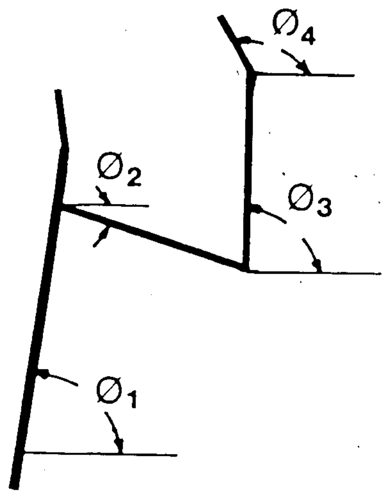

Figure 1: Definition of Segmental Angles. (ϕ_1, ϕ_2, ϕ_3, ϕ_4) are the angles formed by the long axis of the trunk, arm, forearm and hand respectively, with the right horizotal.

II. REVIEW OF LITERATURE

Most of the information related to basketball shooting technique was found in books and manuals. Authors were either basketball coaches, or experts in the area of biomechanics. On the other hand, information about specific muscle torques appeared mainly in scientific periodicals.

A. Literature Related to the Projection of the Ball

Shooting in basketball involves the projection of a ball towards a horizontal rim which is located 3.05 m above the ground. Barnes (1972) explained that if the ball could have dropped vertically at a 90 degree angle to the floor it would have the greatest chance of passing through the basket. As the angle of approach is reduced the "opening" to he basket is sharply decreased. In other words, as the angle of approach diminishes, the probability of scoring decreases since the margin of error is reduced (Hay, 1978).

The projected ball is governed by the force of gravity, therefore the trajectory of its flight follows the path of a parabola and is dependent upon its speed and angle at release. Theoretically, there are many possible combinations of speeds and angles of release that will result in a successful shot. Practically, a higher arc (which results in a better angle of approach), demands a greater ball velocity at release, and requires application of a larger force on the ball. This in itself is a disadvantage, since it is known that it is more difficult to control a force imparted by muscles under strain (Causy, 1970). The action of such muscles under strain or extreme tension is uncontrollable and unpredictable (Bunn, 1972). Furthermore, the higher the arc, the more important control becomes, since any deviation from the desired direction is magnified when an object travels further (Broer, 1973). Barnes (1972) summarized this issue, saying that a lower trajectory allows somewhat better control of velocity and better accuracy, but the ball "sees" a smaller rim opening. On the other hand, a higher trajectory requires increased velocity, and decreases the accuracy despite the larger apparent target. Consequently a compromise is required. Barnes suggested that :

> ...by stating that the optimum trajectory is that which results in the ball entering the basket at the highest possible angle commensurate with the lowest possible velocity we shall have said all that can be said dogmatically about trajectory.

Coleman (1975) suggested that:

> ...the optimum arc will be the one which enables the ball to enter the rim at a position as near vertical as possible proportional to the lowest possible velocity of release.

All factors considered, Mortimer (1951) suggested that projection of the ball with a 58 degree angle of release, in combination with the velocity necessary to put the ball through the center of the basket, allows the shooter the greatest margin of error. Sharman (1965) advocated a medium arc which resulted from an angle of release between 35 and 45 degrees. Hartley and Funton (1971) suggested that the medium arc would have some of the advantages of both the high and the low arc and is the one that most players should use. However, the researchers suggested that the angle of release that would result in a medium arc is somewhere between 55 and 60 degrees. Barnes (1972) suggested that it is best to shoot the ball at a 45 degree angle of release. Hay (1978) suggested that in the case of a 15' shot released at a height of 7', an angle of release between 49 and 55 degrees is likely to provide the shooter with a greater chance of success than any angle outside this range. In an experimental study, Gorton (1978) found that female basketball players shot the ball with a mean angle of release of 48.94 degrees and males, 39.05 degrees. Hudson (1983) found in his study that the mean angle of release used by a group of highly skilled basketball players was 52 degrees and the range was 46-60 degrees.

B. Description of the One hand Set Shot

In the preparation for the shot the feet are positioned behind the free throw line. The position of the feet is a matter of personal preference, although most players prefer the foot under the shooting arm to be slightly forwards (Cousy, 1970). The body is slightly crouched with the knee and hip slightly bent, the joints relaxed and the head up directly above the mid-point between the feet (Wooden,1980). The elbow of the shooting arm should be aligned directly under the ball, and should point towards the basket (Wooden and Sharman 1975). The shooting hand is positioned on the low back side of the ball, with the fingers evenly and comfortably spread behind and under the ball with the palm facing the basket (Macauley, 1971; Hartley and Funton, 1971; Cousy, 1970). The non-shooting hand is placed on the side of the ball and it is used to provide additional support, balance and control. The level that the ball is carried when in the set position varies from player to player (Cousy, 1970). Some players initiate the movement from chest height, while others prefer to begin the shooting action from a higher position in front of and above their eyes.

The shot is initiated with the movement of the upper limb into a preparatory position accomplished by shoulder and elbow flexion. In the propulsive phase the knees and hips are extended, the shoulder is flexed, the elbow is extended and the wrist is flexed (Wooden, 1980; Cooper and Siedentop, 1975; Buckley, 1962). The wrist action is a vital part of both the power and guiding action of the shot. As the wrist snaps and flexes forward, downward and slightly inward, additional force is developed (Macauly, 1971; Hartley and Funton, 1971). The wrist action provides major control for the application of force on the ball (Buckley, 1962). It is assisted in varying degrees by the fingers and finger tips that pass under the ball and impart backward spin to the ball (Hartley and Funton, 1971). The non-shooting hand supports the ball during its upward motion, but it loses contact with the ball prior to the moment of release and thus the impetus to the ball is transmitted solely by the shooting hand (Cousy, 1970). Following the release the shooting hand should follow through towards the basket. Hartley and Funton suggested the wrist be left in a snapped position out in front of the head. The elbow should be kept relatively straight but not stiff.

C. Shooting in Wheelchair Basketball

Owen (1982) suggested that the one hand push shot (set shot) is the most commonly used shot in wheelchair basketball. Skillen (1983) emphasized that although most players automatically start shooting with two hands to compensate for the lack of power from the lower extremity, the proper one hand set shot should be taught, and the two hand shooting technique should be discouraged.

Description of the Wheelchaired One Hand Set Shot

For a proper execution of the one hand set shot the wheelchair should be pointed at the basket (Owen 1982). Body balance should be maintained and the head and the trunk should be kept as straight and erect as possible. During preparation for the shot, the elbow of the shooting arm should be directly under the ball, pointing towards the basket. The wrist should be cocked and the ball resting on the fingers which are comfortably spread under and behind the ball (Owen, 1982; Shavor, 1981).

The shooting action should be smooth and sequential (Skillen, 1983). As it starts the shoulder is flexed and the elbow is extended with the forearm moving forward away from the body and downward. The wrist and fingers are snapped and add power to the shot. A complete and even exaggerated follow through is suggested (Owen, 1982).

The shot should not be modified unless the athletes physical disability prevents execution along this line (Skillen, 1983). Considering the angle of release, Owen (1982) suggested that it is important to use enough arc on the ball since wheelchair players shoot the ball from a position two to six feet lower than able-bodied players. Owen suggested that wheelchair players should use a minimum angle of release of 45 degrees.

D. Literature Related to the Study of Segmental Contribution

Efforts to further understand the mechanical basis of skilled motor performance have led researchers to question the role of individual body segment, or more accurately, the contribution of muscles acting upon a specific joint. Two methods have been developed to approach the problem. The first is based upon joint immobilization and measurement of external performance; while the second involves computation of the internal muscle forces and torques responsible for the performance (Miller, 1980).

The Joint Immobilization Technique

When the joint immobilization method is used, the subject executes a given skill and the criterion of performance is recorded. The individual is then restrained in some way in an attempt to eliminate the influence of movement at a particular joint or joints. Under these conditions the subject attempts to perform the original skill. Decrement in the value of the performance criterion is taken to be an index of the role of the immobilized segment (Miller, 1980).

The joint immobilization technique was employed by different researchers who studied segmental contribution in several skills. Luhtanen and Komi (1978) and Payne et al. (1968) studied the contribution of different segments in vertical jumping. Broer and Zernicke (1979), Hoshikawa and Toyoshima (1976) and Toyoshima et al. (1974) investigated throwing performance. Lanaue (1936, as cited by Miller, 1980) used the immobilization method to study the contribution of different segments in spring board diving, and Davis and Blanksby (1976, as cited by Miller, 1980) immobilized segments to study their contributions in bowling.

The Resultant Muscle Torques Technique

The second method focuses upon resultant muscle torques and thus requires kinetic analysis (Miller, 1980).

Kinetics is that section of dynamics concerned with the forces initiating and altering motion (Miller and Nelson, 1973). Using kinetic analysis, one tries to relate observable configurations or motions of the body to the forces which must be acting in order to maintain those configurations or produce those motions (Andrews, 1974). Such an analysis is considered to be the highest level of mechanical analysis and holds the greatest promise for increasing our understanding of the intricacies of human motion (Miller and Nelson, 1973).

Traditionally, kinetic analysis was oversimpliplified, and the athlete was treated as a particle (Miller and Nelson, 1973). More recently, mathematical models of varying sophistication have been developed and used in conjunction with other measurement tools, and have provided a fruitful approach to the quantitative assessment of human motion (Miller and Nelson, 1973).

Kinetic analysis is based upon, seven essential steps (Plagenhoef, 1966):

1. Determination of segment length
2. Determination of segment weight
3. Determination of segment CM location, and length of radius of gyration
4. Motion photography

5. Motion tracing
6. Determination of instantaneous linear and angular velocities and accelerations of each segment along the time interval to be studied
7. Determination of joint forces and resultant muscle torques.

The first four steps are concerned with estimation of body segment parameters. Several methods have been proposed and used in the past for the determination of body segment parameters (Dempster, 1955; Barter, 1957; Witsett, 1963; Hanavan, 1964; Clauser, 1969; Jensen, 1976; Hatze, 1980). As yet, no single technique has gained universal acceptance (Miller and Nelson, 1973).

Of all segmental properties required for kinetic analysis, it is most difficult to estimate the moment of inertia. Miller (1980) suggested that in many studies this parameter is calculated based upon segmental mass and radius of gyration. The radius of gyration is approximated as a function of segment length.

In steps 5 and 6 cinematography and data reduction techniques are used to obtain kinematic parameters of the performance through the use of motion-picture analysis (Miller and Nelson, 1973; Northrip et al., 1983).

In the final step, all the parameters obtained from the preliminary stages (step 1-6) are used in the equations of motion for the ultimate determination of joint forces and muscle torques. For this purpose, body segments are realistically idealized as rigid bodies (Andrews, 1974). Andrews explained that the bony internal structure of any anatomical segment conforms closely to the idealization of a rigid body. The surrounding tissues, although deformable, usually undergo relatively small changes in size and/or shape during segment motion. These factors, coupled with the mathematical simplicity associated with the dynamic analysis of rigid bodies, as opposed to deformable bodies, have been used to justify the representation of body segments as rigid bodies for the purpose of biomechanical kinetic analysis.

The equations of motion used for the kinetic analysis are based on either force-mass-acceleration relationships, work energy principles, or the impulse momentum relationship (Miller and Nelson, 1973).

Kinetic Analysis Studies of Sports Performance

Fenn (1930) pioneered kinetic description of air phase locomotion. Elfman (1940) expanded upon Fenn's effort by studying the muscle moments generated about different joints during running. The researcher reported 4.5 Kgm maximum shoulder extension torque, 2.3 Kgm maximum elbow flexion torque, 2.6 Kgm maximum elbow extension torque, 14.1 Kgm maximum hip flexion torque, 11.1 Kgm maximum hip extension torque, 19.7 Kgm maximum knee flexion torque, 10.0 Kgm maximum knee extension torque, 18.2 Kgm maximum ankle extension torque and 13.5 Kgm maximum ankle flexion torque. Plagenhoef (1968a) presented two computer programs for kinetic analysis. Plagenhoef (1968b) used his programs to study the moments of force generated at the lower extremities during running (two subjects at 5.8 m/s pace). Plagenhoef reported maximum hip flexion torques of approximately 810 Nm. Dillman (1971) studied the kinetics of the lower limb during the recovery phase in running. Five highly skilled runners were subjects in this study (average velocity 9.2 m/s). Dillman reported maximum hip flexion torques of 392 Nm . Ariel (1974) studied knee joint torques during deep knee bends with weights. Ariel found a maximum torque of 250 Nm. The author didn't specify whether this was flexion or extension torque. Zernicke (1974), Roberts (1974) and Zernicke and Roberts (1976, 1978) studied the relative contribution of kicking limb segments. Average maximum hip flexion torques were found to be 119, 120 and 271 Nm for slow, medium and high speed kicking velocities, respectively. Average maximum torques of 122 Nm were calculated for the knee extensors, and 20 Nm torques were calculated for the dorsi flexors at the ankle joint. This report concentrated upon sequential patterns of torque development and also included a validation experiment. The researchers found that the technique (two dimensional model) produced reasonable estimates of both magnitude and temporal sequencing of the kinetic variables of interest. Mean percent agreement between vertical ground reaction forces computed from the modeling technique compared to those forces simultaneously recorded by a force platform was greater than 95 (Zernicke et al., 1976). Garhammer (1976) developed a five link model to study net joint forces and muscle torques during the pull phase of a snatch weightlifting performance. Subjects in this study were international caliber weightlifters. Garhammer found that the lightest weightlifter (52 Kg) generated 250 Nm hip extension torques and 150 Nm knee extension torques whereas the heaviest weightlifter generated 600 Nm hip extension torques and 200 Nm knee extension torques. Garhammer validated his model by filming a similar performance on a force platform. The researcher found that the calculated and directly measured forces differed by less than 10 percent during most of the pulling movement. Cavanagh et al. (1977) studied the kinetics of long distance running using 22 outstanding athletes running on a treadmill at 5 m/s. The investigator reported on maximum hip flexion torques of only 25 Nm .

Mann (1981) conducted a kinetic analysis of sprinting performance. The researcher focused upon muscle torque patterns about the ankle, knee, hip, elbow and shoulder joints, but didn't quantify maximum torque values. Dessureault (1977) studied the kinetics of shot-putting. The researcher compared the force and torque development patterns of athletes of different skill levels. Analysis concentrated upon observation of force-time curves, but quantification of maximum torque values was not presented.

Kinetic Analysis Studies on Basketball Shooting

Research in basketball shooting in the past has been primarily concentrated upon selected kinematic parameters. Kinetic analyses of basketball shooting performances are rare. Gorton (1978) used recorded data from a force platform to study the horizontal and vertical ground reaction forces exerted during the final two strides and at take-off in basketball jump shots. No information was found on studies that scientifically investigated the contribution of different body segments in basketball shooting or in wheelchair basketball shooting.

Values for maximum muscle torques were published by Plagenhoef (1971). For basketball set shooting, Plagenhoef reported on maximum torque values of 0.5 Kgm for shoulder extension, 1.1 Kgm for elbow flexion, 2.7 Kgm for elbow extension, 0.42 Kgm for wrist flexion and 0.03 for wrist extension. For basketball jump shooting, Plagenhoef suggested maximum torque values of 4.1 Kgm for shoulder extension, 1.5 Kgm for elbow flexion, 3.0 Kgm for elbow extension, 0.6 Kgm for wrist flexion and 0.25 Kgm for wrist extension. No information was given as to the method undertaken to approximate the reported values.

III. METHODOLOGY

> ....It is vital that we realize that instrumentation and methodology are not ends in themselves. With equipment no more complicated than steel balls, planks of wood and water clocks, Gallileo performed experiments that were fundamental to our view of mechanics. In the face of a contribution that represented so much with so little experimental sophistication it is a challenge to us to contribute a little with so much.... .
>
> (Cavanagh, 1975).

This chapter deals with the methods and techniques undertaken to collect, reduce and analyze the data in this study.

A. Subjects

Thirty male athletes were subjects for this study. Ten subjects were stand-up basketball players, members of two Canadian intercollegiate teams. Twenty subjects were physically disabled wheelchair basketball players. Among the wheelchair athletes 10 subjects were classified as class II athletes, and the remainder (n=10) were class III physically disabled wheelchair basketball players.

B. Data Collection

A photo Sonics 1PL 16 mm pin registered camera was positioned on a tripod 18 meters from the performer, with its optical axis perpendicular to the plane of action. The camera was loaded with Kodak 7250 film (ASA 400) and was set to operate at 150 frames per second for all trials. The camera was operated with a shutter angle of 45 degrees and with an exposure time of 1/1200 second. Frame rate was calibrated using a Photo Sonics electronic internal timing light generator at a frequency of 10 Hz. Filming sessions were conducted on four different occasions. Performances of wheelchair free throws were filmed during the half time intermissions and between games in two international wheelchair basketball tournaments, whereas performances of stand-up free throws were filmed during practice sessions of the University of Alberta and University of Victoria intercollegiate basketball teams. Prior to each filming session, a reference measure 1 meter long was filmed. The ratio between the projected image size and the actual size of this measure was the conversion factor used to change film measurements to real life dimensions.

C. Data Reduction

Film analysis was accomplished through the use of an electronic digitizing system. The film was projected onto a Bendix Digitizing Board by a Triad pin registered film analyzer. Body segment end points as well as five points on the circumference of the ball were digitized and stored on magnetic tapes through the use of the Hewlett Packard 9864A Digitizer and Hewlett Packard 9825B micro computer. The film analyzer was levelled and positioned perpendicular to the digitizing surface in order to assure proper film alignment. The Human Scale Anatomical data (Diffrient, 1979) was used as a guideline to locate segment end points for digitizing purposes.

D. Data Analysis

Data obtained from the film consisted of X, Y position coordinates. Since there is always a certain degree of error associated with the data reduction procedure, the raw data were smoothed using second order Butterworth digital filtering technique Cut-off frequencies of 10, 10, 12, 15, 15 Hz were used to smooth the raw data of the distal trunk, proximal arm, proximal forearm, proximal hand and distal hand, respectively. Data from the ball-hand contact point were smoothed with a 15 Hz cut-off frequency, and the X, Y cartesian coordinates of the calculated center of mass of the ball were smoothed using a 10 Hz cut-off frequency (Patrick et al., 1980).

Calculation of Distances, Angles, Velocities and Accelerations of Body Segments

Kinematic parameters were calculated from smoothed X, Y cartesian coordinates.

Segmental angles were formed by the long axis of each segment and the right horizontal, and calculated as follows (Winter, 1979).

$$\theta j = \arctan\left(\frac{Yd - Yp}{Xd - Xp}\right)$$

where:

θj = segmental angle for segment j

Xp , Yp = cartesian coordinates of the proximal end point of segment j

Xd , Yd = cartesian coordinates of the distal end point of segment j

Linear and angular accelerations were calculated using central differences (Miller and Nelson, 1973).

$$a_{x\,i} = \frac{X_{i+1} - 2X_i + X_{i-1}}{(t)^2}$$

$$a_{y\,i} = \frac{Y_{i+1} - 2Y_i + Y_{i-1}}{(t)^2}$$

$$\alpha_i = \frac{\phi_{i+1} - 2\phi_i + \phi_{i-1}}{(t)^2}$$

where:

$a_{x\,i}$ = average horizontal acceleration of a point between frames i-1 and i+1

$a_{y\,i}$ = average vertical acceleration of a point between frames i-1 and i+1

α_i = average angular acceleration of a point between frames i-1 and i+1

t = time interval between adjacent frames.

ϕ = segmental angle at frame i

X ; Y = cartesian coordinates of a point

Calculation of Selected Kinematic Parameters of the Ball

The center of mass of the ball was approximated based on five digitized points (Appendix B). The X,Y components of the linear acceleration of the ball's center of mass were calculated using finite differences (Miller and Nelson, 1973). The angle of release was found to be:

$$\theta r = \arctan\left(\frac{Yr_{+1} - Yr}{Xr_{+1} - Xr}\right)$$

where:

θr = angle of release

Xr_{+1} = X coordinate of ball's CM one frame after release

Xr = X coordinate of ball's CM at release

Yr_{+1} = Y coordinate of ball's CM one frame after release

Yr = Y coordinate of ball CM at release

The velocity of release was found to be:

$$Vr = \sqrt{\left(\frac{Xr_{+1} - Xr_{-1}}{2t}\right)^2 + \left(\frac{Yr_{+1} - Yr_{-1}}{\quad}\right)^2}$$

where:

Vr = Velocity of release

Xr_{+1} = X coordinate of ball's CM one frame after release

Xr_{-1} = X coordinate of ball's CM one frame prior to release

Yr_{+1} = Y coordinate of ball's CM one frame after release

Yr_{-1} = Y coordinate of ball's CM one frame prior to release

t = time interval between adjacent frames

The angle of approach was predicted based upon an equation derived by Mortimer (1951).

$$\tan(\theta)a = \tan(\theta r) - 2h/d$$

where:

θr = angle of release

θa = angle of approach

h = height of release

d = horizontal distance from the point of release to the center of the rim

Calculation of Resultant Muscle Torques

Kinetic analysis was administered based exclusively on biomechanics cinematography. Accelerations were obtained from kinematic analysis. Segmental lengths were obtained from film analysis. Predicted Body weight, was estimated after accounting for disproportionalities due to amputation or lesion. The estimated weight of the missing segments was added to the body weight of the amputee athletes. The body weight of the paraplegic and post polio disabled athletes was increased 22 percent to account for the atrophied segments. (Appendix D). Segmental weights were then approximated as percentages of predicted body weight (Table 1). Segment centers of mass and radii of gyration were approximated as percentages of segment length (Table 1), and segment moments of inertia were calculated based on segment mass and radii of gyration.

$$Ij = mj * kj$$

where:

Ij = moment of inertia of segment j about a transverse axis through its CM

mj = mass of segment j

kj = radius of gyration of segment j.

Table 1

Anthropometric Data - (after Dempster 1955, as cited by Winter 1979)

Segment	Segment Weight/ Body Weight	Center of Mass/ Segment Length (from proximal end)	Radius of Gyration/ Segment Length (about CM)
Hand	.006	.506	.297
Forearm	.016	.430	.303
Arm	.028	.436	.322

For the kinetic analysis, the upper limb and the ball were represented as a system of four homogeneous rigid bodies. Each was considered individually in a sequential analysis which began with the ball and progressed to the hand, forearm and arm.

The free body diagrams of the segments (Figure 2 through figure 5) show segmental weights directed vertically downward from their respective mass centers, joint reaction forces represented by horizontal and vertical components, and the resultant muscle torque crossing each joint. The force-mass-acceleration method of deriving the equations of motion for rigid bodies (Miller and Nelson, 1973) was employed to determine the joint reaction forces and resultant muscle torques at successive instances in the movement.

The following equations of motion were used.

$$mb * a_{x5} = R_{x4}$$
$$mb * a_{y5} = R_{y4} - 9.81mb$$

$$m_4 * a_{x4} = R_{x3} - R_{x4}$$
$$m_4 * a_{y4} = R_{y3} - R_{y4} - 9.81m_4$$
$$I_4 * \alpha_4 = T_3 + R_{x3} * .506L_4 * \sin(180-\phi_3) + R_{y3} * .506L_4 * \cos(180-\phi_3) + R_{x4} * (L_5 - .506L_4) * \sin(180-\phi_3) + R_{y4} * (L_5 - .506L_4) * \cos(180-\phi_3)$$

$$m_3 * a_{x3} = R_{x2} - R_{x3}$$
$$m_3 * a_{y3} = R_{y2} - R_{y3} - 9.81m_3$$
$$I_3 * \alpha_3 = T_2 - T_3 + R_{x3} * .57L_3 * \sin(180-\phi_2) + R_{y3} * .57L_3 * \cos(180-\phi_2) + R_{x2} * .43L_3 * \sin(180-\phi_2) + R_{y3} * .43L_3 * \cos(180-\phi_2)$$

$m_2 * a_{x2} = R_{x1} - R_{x2}$
$m_2 * a_{y2} = R_{y1} - R_{y2} - 9.81m_2$
$I_2 * \alpha_2 = T_1 - T_2 + R_{x2} * .564L_2 * \sin(180-\phi_1) + R_{y2} * .564L_2 * \cos(180-\phi_1) + R_{y1} * .436L_2 * \sin(180-\phi_1) + R_{y1} * .436L_2 * \cos(180-\phi_1)$

where:

T_4 = resultant muscle torque at the wrist joint

T_3 = resultant muscle torque at the elbow joint

T_2 = resultant muscle torque at the shoulder

R_{x4} ; R_{y4} = X,Y components of the reaction force at the wrist joint

R_{x3} ; R_{y3} = X,Y components of the reaction force at the elbow joint

R_{x2} ; R_{y2} = X,Y components of the reaction force at the shoulder

$mb;m_4;m_3;m_2$ = mass of ball, hand, forearm and arm respectively

$I_4;I_3;I_2$ = moments of inertia of the hand forearm and arm respectively, about a transverse axis through segment's CM

$\phi_4;\phi_3;\phi_2$ = angle of the hand,forearm and arm respectively

$L_4;L_3;L_2$ = length of hand, forearm and arm respectively

L_5 = distance between ball-hand contact and proximal hand end point

$\alpha_4;\alpha_3;\alpha_2$ = angular acceleration of the hand, forearm and arm respectively

a_{x5}; a_{x4}; a_{x3}; a_{x2} = horizontal acceleration of the ball, hand
forearm and arm respectively

a_{y5}; a_{y4}; a_{y3}; a_{y2} = vertical acceleration of the ball, hand
forearm and arm respectively

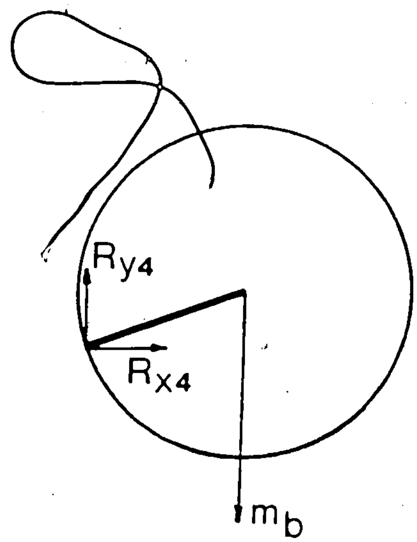

Figure 2: Free Body Diagram of the Ball

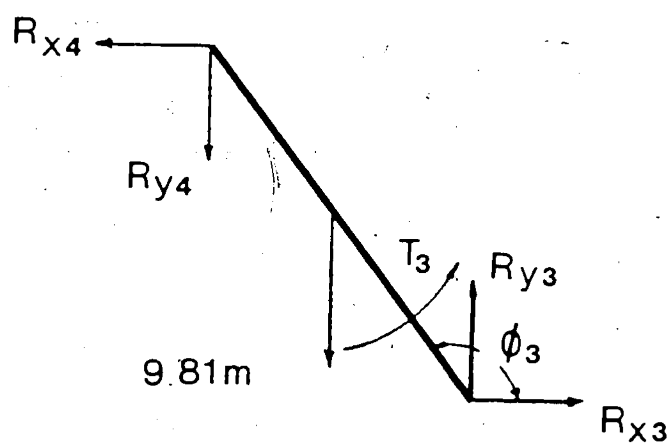

Figure 3: Free Body Diagram of the Hand

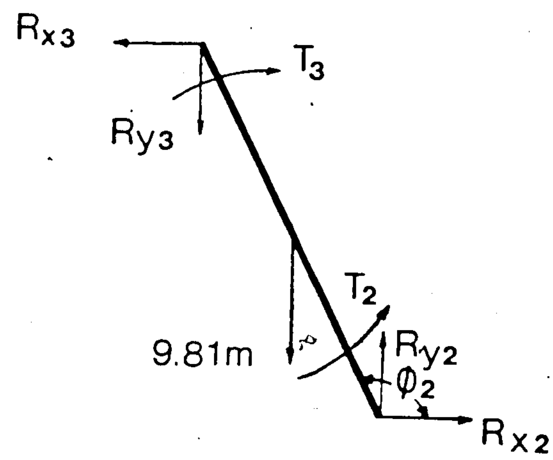

Figure 4: Free Body Diagram of the Forearm

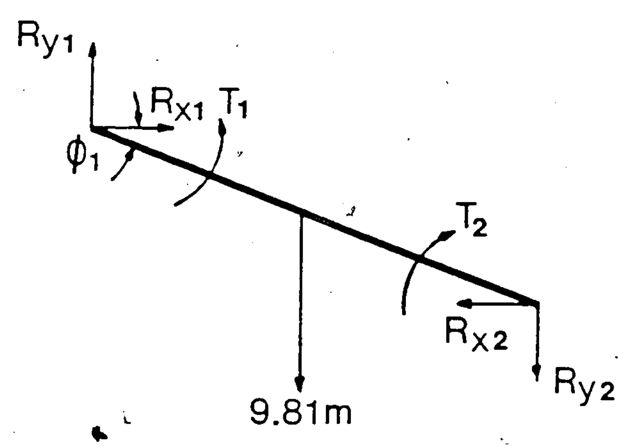

Figure 5: Free Body Diagram of the Arm

E. Statistical Procedures

The following statistical procedure was administered:

1. One way ANOVA was used to identify over all differences between means. A .05 level of significance was deemed acceptable.
2. Post-hoc Tukey and Scheffee tests were used to identify differences between means of the subsets under investigation (Ferguson, 1981).

The above procedure was repeated seven times to analyze the results of the following seven variables under investigation:

1. angle of release
2. speed of release
3. angle of approach
4. maximum shoulder flexion torque
5. maximum shoulder extension torque
6. maximum elbow extension torque
7. maximum wrist flexion torque.

The Statistical Package for the Social Sciences (SPSS) was used for the statistical analysis (Nie et al., 1975).

The comparison of relative contributions of different body segments of different groups of athletes to the total force applied to the ball, was based on the contribution of the segment expressed as a percentage of the total force.

IV. RESULTS AND DISCUSSION

The results obtained in this study are presented and discussed in this chapter. The presentation and discussion focus upon:

A. A kinematic analysis of the trajectory of the ball.

B. A kinetic analysis of the torques responsible for the propulsive forces imparted on the ball.

Both the kinematic and the kinetic analyses include a comparative analysis of:

1. The performance of stand-up vs. wheelchair basketball players.
2. The performance of class II vs. class III wheelchair basketball players.

A. Kinematic Analysis of the Trajectory of the Ball

Results

In this study, the trajectory of the ball was examined first. The results are summerized in table 2 . A comparison between stand-up and wheelchair basketball players showed that stand-up basketball players propelled the ball towards the basket with a significantly smaller angle of release ($P < .01$) and a significantly smaller speed of release (P

<.01). Stand-up basketball players released the ball with an angle of release equal to 51.7±3.0 degrees, and a release speed of 6.5±0.4 m/s . Wheelchair basketball players released the ball with an angle of release of 55.5±2.4 degrees, and with a release speed of 7.2±.3 m/s (Figures 6, 7).

Balls projected by stand up basketball players approached the rim with a slightly greater angle of approach (45.9±4.0 degrees) than did balls projected by wheelchair athletes (42.6±3.8 degrees),(Figure 8). The difference between the above values was found significant at .05 significance level, but not at .01 significance level.

A comparison between the two groups of wheelchair basketball players showed that both class II and class III wheelchair basketball players propelled the ball using almost identical speeds of release (7.17±0.4 and 7.15±0.1 respectively) as well as almost identical angles of release. (55.6±2.1 and 55.6±2.8 degrees respectively), (figures 9 and 10). However, the balls approached the rim with a similar but not identical angle of approach. Balls projected by class II wheelchair basketball players approached the rim with an angle of 42.2±3.76 degrees, whereas balls projected by class III athletes aproached the rim with an angle of 43.0±4.0 degrees (figure 11).

Table 2

The Trajectory of The Ball

Parameter	Stand-up	Class II Wheelchair	Class III Wheelchair
Angle of Release (deg.)	51.7±3.0	55.6±2.1	55.6±2.8
Speed of Release (m/sec.)	6.50±0.4	7.17±0.4	7.15±0.1
Angle of Approach (deg.)	45.9±4.0	42.2±3.8	43.0±4.0

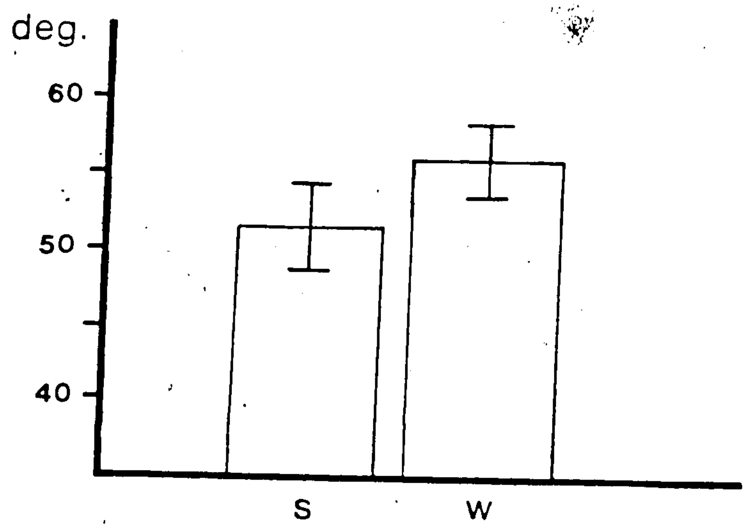

Figure 6:Angle of Release of
Stand-up (S) and Wheelchair (W) Basketball Players

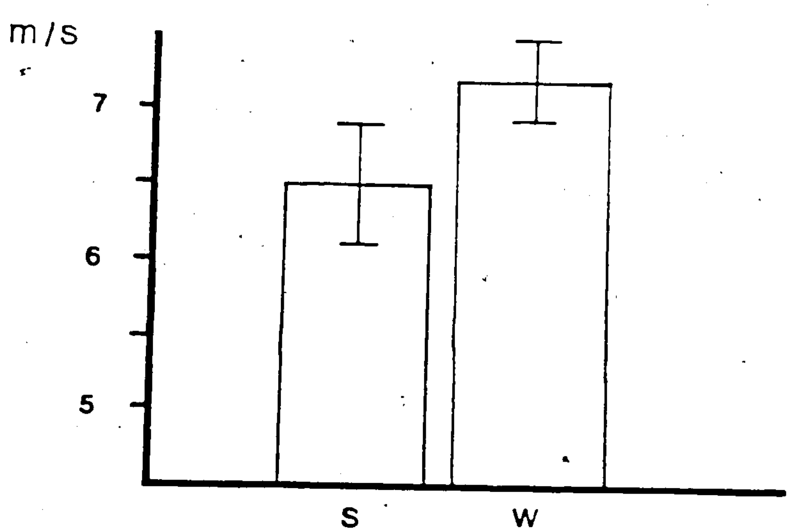

Figure 7: Speed of Release of
Stand-up (S) and Wheelchair (W) Basketball Players

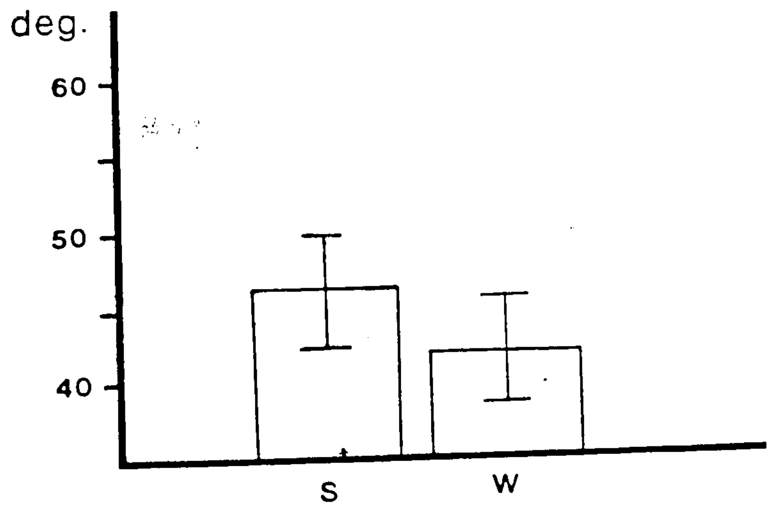

Figure 8: Angle of Approach of
Stand-up (S) and Wheelchair (W) Basketball Players

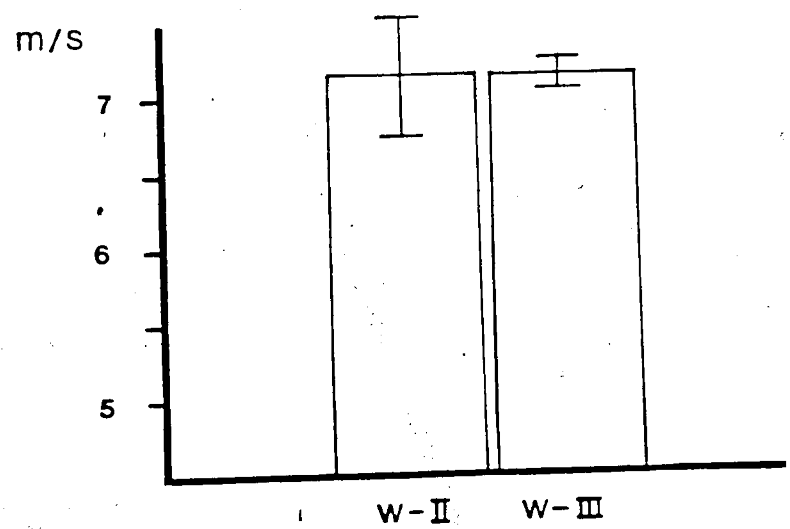

Figure 9: Speed of Release of
Class II and Class III Wheelchair Basketball Players

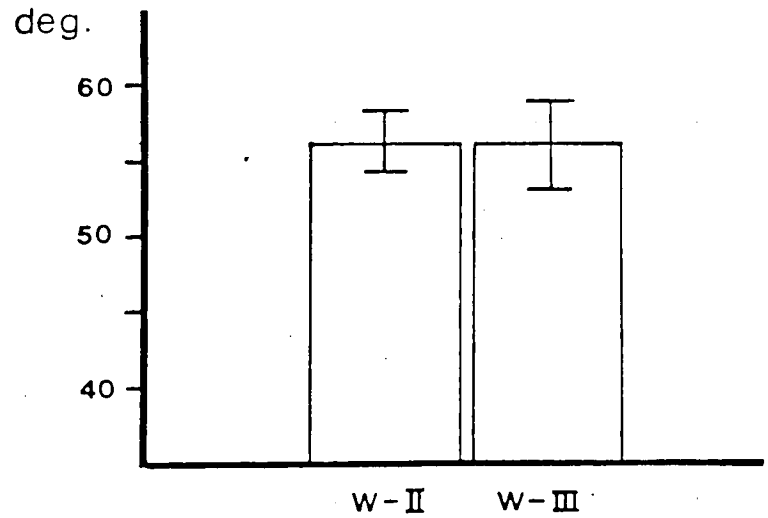

Figure 10: Angle of Release of
Class II and Class III Wheelchair Basketball Players

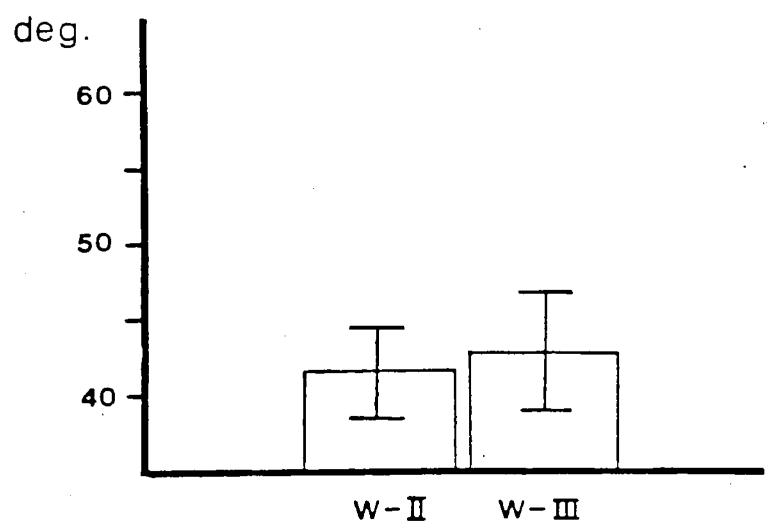

Figure 11:Angle of Approach of
Class II and Class III Wheelchair Basketball Players

Discussion

In this study, stand-up basketball players shot the ball with an angle of release of 51.7±3.0 degrees. Among suggested angles of release, the results of this study support Hay (1978) who suggested that the optimum angle of release from the free throw line lies between 49 and 54 degrees.

Wheelchair basketball players, in this study, propelled the ball with an angle of release of 55.5±2.5 degrees. The results of this study disagree with Owen (1982), who suggested that wheelchair basketball players should use a minimum 45 degree angle of release.

The author of the present study suggests that wheelchair basketball players should use a minimum 50 rather than 45 degree angle of release. A smaller angle of release will usually result in an angle of approach less than 38 degrees. This will dramatically decrease the margin of error and the probability of scoring (Hay, 1978).

The ultimate purpose in free throw shooting is to increase the angle at which the ball approaches the rim and thus to increase the probability of scoring, commensurate with the lowest possible velocity of release (Barnes, 1972).

Analysis of the trajectory of the ball revealed that shots taken by both groups of wheelchair basketball players resulted in a slightly smaller angle of approach compared to stand-up players. A smaller approach angle yields a lesser probability of scoring, despite the significantly greater speed and angle of release. It appeared that both class II and class III wheelchair basketball players chose to use a greater speed of release and as a result placed increased demands on their muscles. Nevertheless, they were unable to achieve a probability of scoring comparable to that achieved by stand-up basketball players.

The close similarity in the trajectory of the ball for both class II and class III basketball players might indicate that when shooting from the free throw line, the increased vertical distance between the point of release and the target for the wheelchair group is the main factor responsible for a disadvantageous trajectory.

The minor differences that were found between the angles of approach for the two wheelchair groups, indicated that class II wheelchair athletes released the ball from a slightly lower height. This might be attributed to the smaller body segment dimensions of class II athletes in this study.

B. Kinetic Analysis

Body Segment Parameters

Individual body segment parameters were estimated and presented in Appendix C. The mean values of the groups' body segment parameters were summarized in table 2. Compared to other groups in this study, Class II wheelchair basketball players had slightly shorter arms (.267 m), forearms (.261 m) and hands (.169 m). They also had lighter arms (2.199 Kg), forearms (1.255 Kg) and hands (.470 Kg). Consequently, the upper limbs of class II wheelchair basketball players in this study had smaller moments of inertia. Their arms, forearms and hands had moments of inertia of .0165, .0079 and .0012 Kgm^2 respectively.

Upper limb segments of class III wheelchair basketball players and stand-up basketball players in this study were of almost identical length. Class III wheelchair athletes' arms, forearms and hands were .280 m, .264 m and .175 m long respectively. Stand-up basketball players had arms .277 m long, forearms .265 m long and hands .175 m long. Examination of segment weights and moments of inertia of both groups revealed that class III basketball players in this study had heavier upper limbs, and thus their upper limbs had greater moments of inertia. Arms, forearms and hands of class III wheelchair athletes massed 2.615, 1.491 and .550 Kg respectively, and had moments of inertia of .0215, .0097 and .0015 Kgm^2 respectively. Arms, forearms and hands of stand-up basketball players massed 2.269, 1.296 and .485 Kg respectively and had moments of inertia of .0182, .0085 and .0013 Kgm^2 respectively.

Table 3

Grouped Body Segment Parameters of Subjects

Segment	Stand-up	Class II Wheelchair	Class III Wheelchair
Arm Length(m)	0.277	0.267	0.280
Weight(Kg)	2.269	2.199	2.615
Moment of Inertia(Kgm^2)	0.0182	0.0165	0.0215
Forearm Length(m)	0.265	0.261	0.264
Weight(Kg)	1.296	1.255	1.491
Moment of Inertia(Kgm^2)	0.0085	0.0079	0.0097
Hand Length(m)	0.175	0.169	0.175
Weight(kg)	0.485	0.470	0.550
Moment of Inertia(Kgm^2)	0.0013	0.0012	0.0015

Results

A kinetic analysis of individual performances, and a statistical analysis conducted to identify differences between means, enabled a comparison to be made between stand-up and wheelchair basketball players, and between class II and class III wheelchair basketball players. The comparison focused upon mean maximum torques generated at the shoulder, elbow and wrist joints of the shooting upper limbs. The results are summerized in table 4.

Wheelchair basketball players in this study generated significantly greater ($p < .05$) maximum shoulder flexion torques (21.2±7.3 Nm) than did stand-up basketball players (15.5±3.6 Nm), (Figure 12).

No significant difference was found between the mean maximum shoulder flexion torques, generated by class II and class III wheelchair basketball players (20.4 ± 7.6 Nm and 22.1 ± 7.3 Nm respectively), (Figure 13).

Figure 14 shows that wheelchair basketball players generated significantly greater ($p < .01$)shoulder extension torques (36.3±11.0 Nm) than did stand-up basketball players (25.0±7.0 Nm).

No significant difference was found between the mean maximum shoulder extension torques generated by class II and by class III wheelchair basketball players. (36.0±13.3 Nm and 36.5±8.8 Nm respectively), (Figure 15).

Wheelchair basketball players generated slightly but not significantly ($p > .1$), greater maximum elbow extension torques (22.8±6.2 Nm) than did stand-up basketball players (20.8±6.3 Nm), (Figure 16).

The difference between the maximum elbow extension torques generated by the two groups of wheelchair athletes was not significant ($p > .1$). Class II wheelchair basketball players generated maximum elbow extension torques of 20.4±4.8 Nm), whereas class III wheelchair basketball players generated equivalent torques of 25.3±6.8 Nm. (Figure 17).

Figure 18 shows that wheelchair basketball players in this study generated significantly greater ($p < .1$) maximum wrist flexion torques (8.7±2.0 Nm) than did stand-up basketball players (7.2 ± 1.6 Nm).

Among the wheelchair athletes, class III wheelchair basketball players generated significantly ($p < .05$) greater maximum wrist flexion torques (9.6±2.0 Nm) than did class II wheelchair basketball players (7.7±1.5 Nm), (Figure 19).

Table 4

Summary of Maximum Muscle Torques (Nm)

Muscle Group	Stand up	Class II Wheelchair	Class III Wheelchair
Shoulder Flexors	15.5±3.6	20.4±7.6	22.1±7.3
Shoulder Extensors	25.0±7.0	36.0±13.3	36.5±8.8
Elbow Extensors	20.8±6.3	20.4±4.8	25.3±6.8
Wrist Flexors	7.2±1.6	7.7±1.5	9.6±2.0

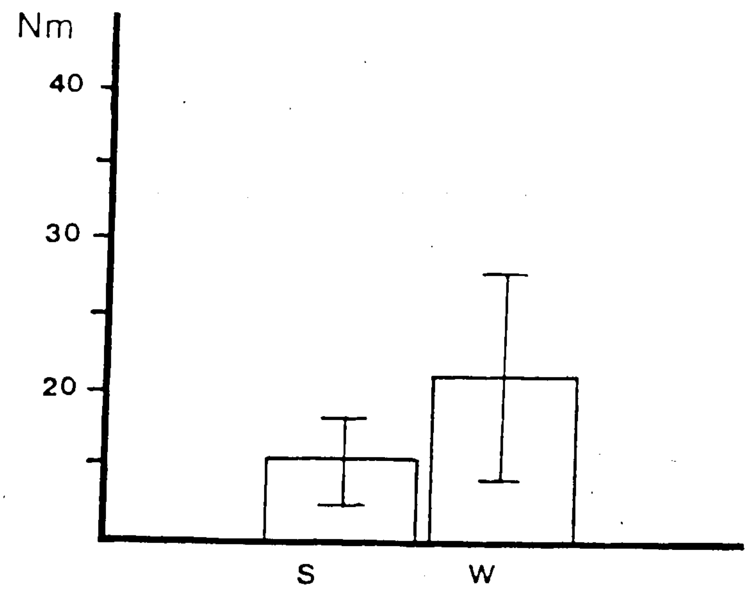

Figure 12: Maximum Shoulder Flexion Tourques Generated by Stand-up (S) and Wheelchair (W) Basketball Players

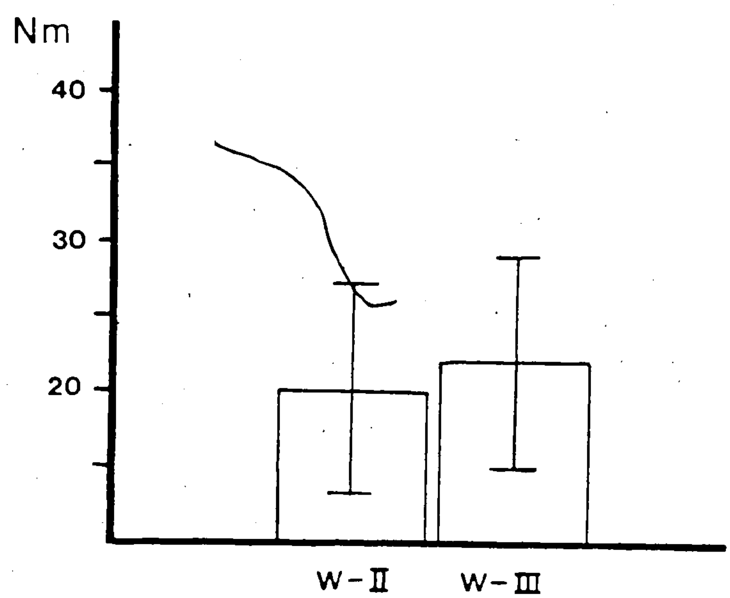

Figure 13: Maximum Shoulder Flexion Torques Generated by Class II and Class III Wheelchair Basketball Players

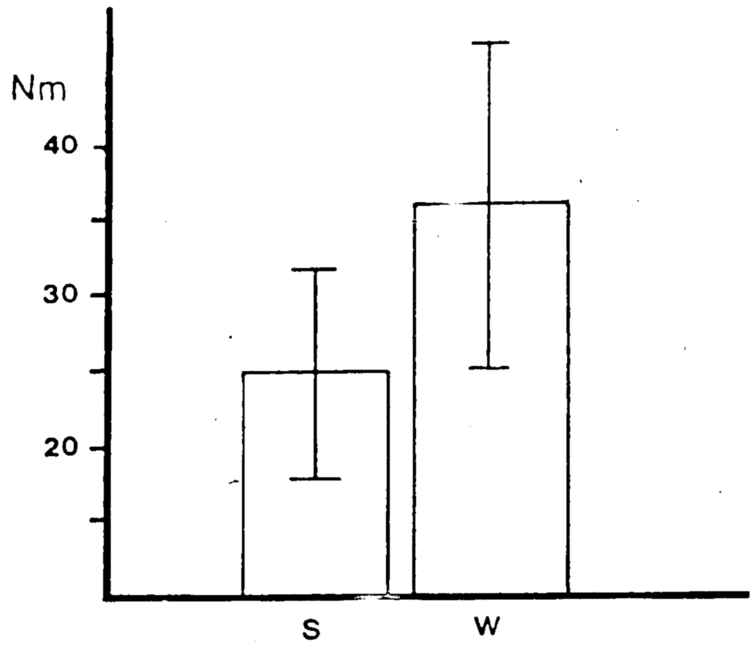

Figure 14: Maximum Shoulder Extension Torques Generated by Stand-up (S) and Wheelchair (W) Basketball Players

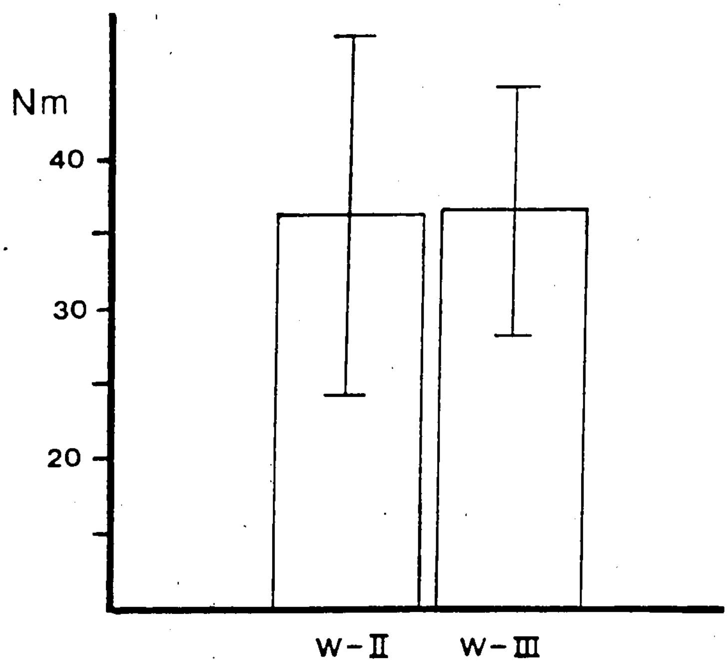

Figure 15:

Maximum Shoulder Extension Torques Generated by Class II and Class III Wheelchair Basketball Players

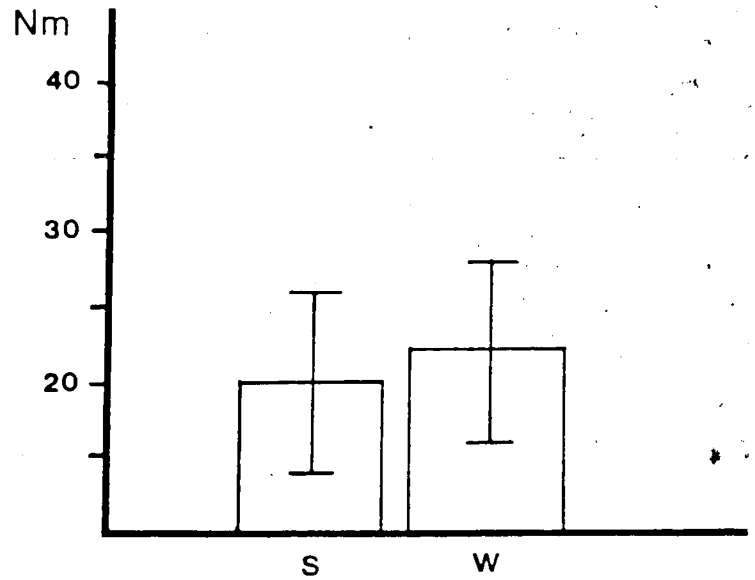

Figure 16: Maximum Elbow Extension Torques Generated by Stand-up (S) and Wheelchair (W) Basketball Players

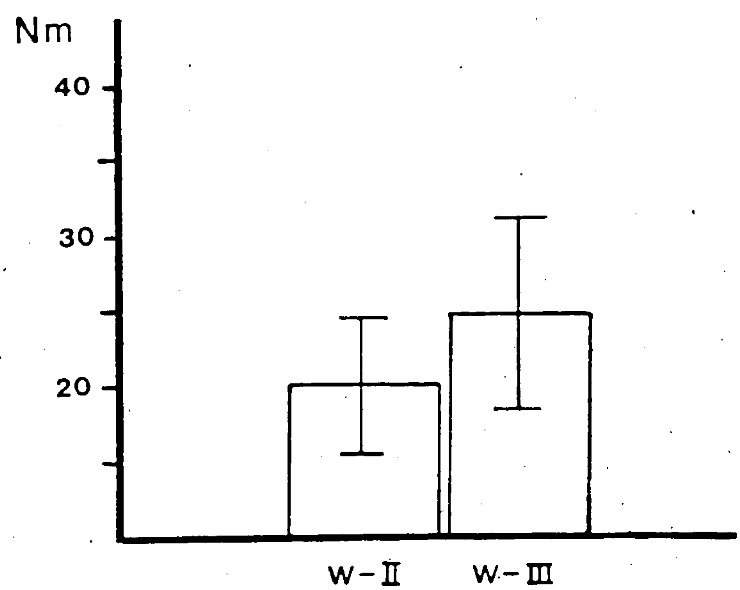

Figure 17: Maximum Elbow Extension Torques Generated by Class II and Class III Wheelchair Basketball Players

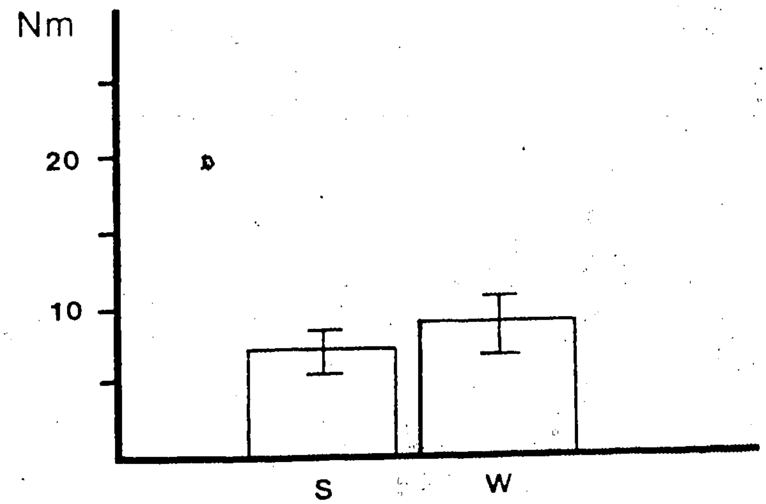

Figure 18: Maximum Wrist Flexion Torques Generated by Stand-up (S) and Wheelchair (W) Basketball Players

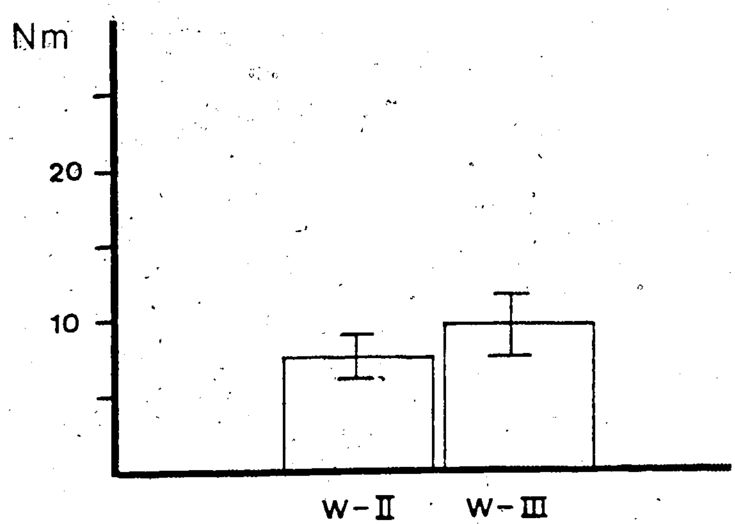

Figure 19: Maximum Wrist Flexion Torques Generated by class II and Class III Wheelchair Basketball Players

Discussion

Within the limitations of this study, it was found that wheelchair basketball players needed greater muscle torques in order to propel the ball towards the basket. However, the increased load was not proportionately distributed amongst the muscle groups involed in the shooting action.

When the performance of wheelchair basketball players was compared to that of stand-up basketball players, it appeared that wheelchair basketball players developed relatively greater shoulder flexion and shoulder extension torques than elbow extension and wrist flexion torques. Numerically, shoulder flexion and shoulder extension torques developed by wheelchair basketball players were greater than those developed by stand-up basketball players by 38 and 45 percent respectively, whereas elbow extension and wrist flexion torques developed by wheelchair basketball players were greater than those developed by stand-up basketball players by only 10 and 20 percent respectively.

Based on the present study the following explanation of the performance of wheelchair basketball players appeared plausible.

The set shooting action in stand-up basketball begins with the extension of the knee and hip joints. The shoulder of the shooting arm is then flexed and further accelerates the ball which is already moving upward. In wheelchair basketball, the lower limbs contribute very little if anything to the shooting action, and thus the shooting upper limb is responsible for accelerating the ball upwards from a stationary position.

It was found in this study that wheelchair basketball players were required to propel the ball with a greater velocity of release and consequently to further accelerate the ball during the propulsive phase. To do so, wheelchair basketball players were required to generate greater resultant muscle torques.

The data led the researcher to hypothesize that wheelchair basketball players optimized their performance by further overloading those muscles responsible for movements of proximal segments. Thus, muscles at the shoulder joint generated most of the increased force required to propel the ball, whereas the additional demands on the elbow extensors and wrist flexors were minimal. Such a minimal increase in force production optimized the conditions for these muscle groups, since they served as ultimate controllers of the direction and the speed of the ball.

A comparison between the two wheelchair groups in this study was carried out with the following two important factors in mind:

1. Class II wheelchair basketball players in this study were both shorter and lighter than class III wheelchair basketball players. Class II wheelchair basketball players had shorter and lighter upper limb segments, and therefore required smaller torques to overcome the inertial resistance of these segments.

2. The kinetic analysis revealed that there were large individual differences among the performances of wheelchair basketball players in this study. It appeared that physically disabled athletes compensated differently for their specific disability.

Given the above, it was suspected that the differences observed between class II and class III wheelchair basketball players could be attributed to the smaller body dimensions of class II wheelchair athletes in this study, and should not be considered as true or typical differences between the general population of class II and class III wheelchair basketball players. To account for the different anthropometric characteristics of the wheelchair groups in this study, the author conducted a comparison that focused upon relative rather than absolute differences.

When the performance of class II wheelchair athletes was compared to that of class III wheelchair basketball players on the basis of relative contribution of different body segments, it was found that class II wheelchair basketball players generated relatively greater shoulder flexion and shoulder extension torques than elbow extension and wrist flexion torques. In other words shoulder flexion and shoulder extension torques developed by class II wheelchair basketball players were 92.2 percent and 98.8 percent respectively, of the equivalent torques developed by class III wheelchair basketball players. Elbow extension and wrist flexion torques of class II wheelchair basketball players were 80.8 percent and 80.0 percent respectively, of the equivalent torques developed by class III wheelchair basketball players.

Assuming that the differences in force production due to differences in body segment parameters were proportional for all segments and for all subjects, it appeared that class II wheelchair basketball players used their shoulder flexors and shoulder extensors relatively more than class III wheelchair basketball players. These data reflect compensation for the lack of contribution of the lower limbs in the case of class II wheelchair athletes, and minimal contribution of the lower limbs to the performance of class III wheelchair athletes. Here again, it was speculated that class II wheelchair basketball players optimized their performance so that segments located further away from the open end of the kinetic chain would contribute relatively more as force producers, whereas the segments located towards the open end of the kinetic chain would perform under less strain. This would enable them to control the speed and direction of the ball prior to and during release.

Research that accounts for the differences in individuals' body segment parameters is needed to substantiate this theory. The researcher would like to suggest that future studies should include larger samples to account for the large variation in the observed criterion within the groups studied. However, the large within group variability did indicate that different wheelchair athletes compensated differently for their physical disability and thus attempts to generalize might lead towards invalid conclusions.

V. SUMMARY AND CONCLUSIONS

The purpose of this study was to analyze and to compare the basketball shooting performance of stand-up basketball players with the basketball shooting performance of wheelchair basketball players of two different levels of disability. Selected kinematic and kinetic factors were examined based exclusively on data obtained from biomechanics cinematography.

Kinematic analysis concentrated upon the trajectory of the ball. Angle of release and speed of release were calculated from film data whereas the ball's angle of approach was predicted using film data and principles of trajectory.

Kinetic analysis concentrated upon resultant muscle torques, with an attempt to understand the role of individual segments in the performance. Individual body segment parameters were estimated, the system was defined, equations of motions were developed and maximum shoulder extension, shoulder flexion, elbow extension and wrist flexion torques were computed.

It was found in this study that when compared to stand-up players, wheelchair basketball players projected the ball towards the basket with a greater speed of release and a higher angle of release. Nevertheless, their shots approached the rim with a smaller angle of approach.

Wheelchair basketball players generated greater overall maximum muscle torques. Specifically, they generated relatively greater maximum shoulder flexion and shoulder extension torques than elbow extension and wrist flexion torques.

Class II and class III wheelchair basketball players used almost identical angles of release and speeds of release. However, balls projected by class II wheelchair basketball players approached the rim with a slightly smaller angle of approach.

Class II and class III wheelchair basketball players in this study generated similar shoulder flexion and shoulder extension torques, but class II wheelchair basketball players generated smaller elbow extension and wrist flexion torques. It was suggested that these results should be carefully interpreted since class II wheelchair athletes in this study had shorter and lighter body segments.

Within the limitations of this study, the obtained results led to the following conclusions:

1. Mechanically, it was more difficult for wheelchair basketball players to shoot from the free throw line than it was for stand-up basketball players because:

   a. The seated position necessitated shooting with a higher angle of release and a greater speed of release.

   b. Wheelchair basketball players employed greater torques to propel the ball towards the basket.

2. The seated position rather than the level of disability

was mainly responsible for the trajectory of the ball employed by wheelchair basketball players.

3. Different segments of the shooting upper limb contributed differently to the performance of class II and class III wheelchair basketball players.
4. The methods employed in this study (i.e. 150 frames/sec cinematography, followed by smoothing with joint specific cut-off frequencies) revealed information that was sufficiently accurate for kinetic analysis.
5. The model that was employed in this study yielded maximum muscle torques that were in agreement with the approximated muscle torques reported by Plagenhoef (1971). Consequently, it was concluded that it is possible to investigate the kinetics of basketball shooting using a model that includes the ball as an integral segment of the system of rigid bodies. The ball-hand contact point of such a model should vary with time.

It is recommended that:

1. The distance from which free throws are shot in wheelchair basketball should be re-evaluated considering the mechanical factors discussed in this paper, as well as relevant sociological and psychological factors.
2. As long as free throws in wheelchair basketball are shot from the present distance, wheelchair basketball players

and coaches should limit their expectations and set realistic goals as to the rate of success one can expect. This might avoid frustration and help coaches in their practice plan.

3. Other modifications of the game of wheelchair basketball should also be considered (i.e. lower baskets, smaller basketballs or 3 attempts for 2 points from the free throw line).
4. This study should be replicated comparing athletes with similar body segment parameters.
5. This study should be replicated, using larger samples, to better identify differences between means.
6. Similar studies that include all classes of wheelchair basketball players should be conducted.
7. Similar studies should be conducted in which the effect of the horizontal distance between the shooter and the basket will be introduced as a second independent variable.
8. Similar studies should be conducted in which temporal patterns of torque development are investigated.
9. Similar studies should be conducted in which subjects are physically able athletes who play both stand-up and wheelchair basketball on a regular basis.
10. Attempts should be made to develop a method to estimate segmental weight for paraplegic and post polio wheelchair athletes.

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VI. APPENDIX A: N.W.B.A Classification System

The game of wheelchair basketball was first played by paraplegic veterans of world war II. By 1949, many teams were formed and disabled athletes of all types of disabilities played togegher regardless of their disabilities (owen, 1982). Over time the game became more competitive and the number of severly disabled paraplegic players dramatically decreased. Consquently, at the 1963 annual meeting, the member teams of the N.W.B.A voted to institute a player classification system in an attempt to encourage all disabled individuals, especially the severely disabled, to participate, and to ensure participation of players with differing levels of disability on every team (owen, 1982).

The player classification system in wheelchair basketball relies upon a medical model to determine the functioning muscles and it places each player in one of three classes. The criteria for each of the classes are as folows (Shavor, 1981).

Class I- Complete motor loss at T-7 or above or comparable disability where there is a total loss of muscular function originating at or above T-7.

Class II- Complete motor loss between T-8 through L-2, where there may be useful motor power of the hips but no useful motor power of the lower extremities. Also included in this class are amputees with bi-lateral hip disarticulation.

Class III- All other Physical disabilities as related to lower extremety paralysis or peresis originating at L-3 or lower. All lower extremety amputees except those with bilateral hip disarticulation.

According to the N.W.B.A rules, at any time no more than three class III players can be on the court, and there cannot be more than 12 points per team on the floor.(class I = 1 point ; class II = 2 points ; class III = 3 points). In international competition the players' points can total only 11.

VII. APPENDIX B: Calculation of Ball s Center of Mass

The following trigonometric principles were employed to locate the ball's CM (figure 20)

if : $AD = BD$

and : $DC \perp AB$

then : The line DC will pass through the centre of the circle.

similarily,

if : $FG = EG$

and : $GM \perp EF$

then : the line GM will pass throuth the centre of the circle

Consequently, the line CD and the line GM will intersect at the centre of the ball.

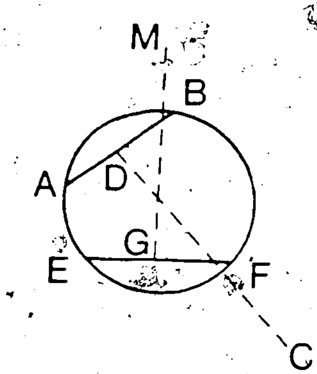

figure 20:
calculation of ball's center of mass.

VIII. APPENDIX C: Individual's Body Segment Parameters

Table 5

Body Segment Parameters of Stand-up Basketball Players

	Arm Length (m)	Forearm Length (m)	Hand Length (m)	Arm Weight (kg)	Forearm Weight (kg)	Hand Weight (kg)	Arm Moment of Inertia (kgm^2)	Forearm Moment of Inertia (kgm^2)	Hand Moment of Inertia (Kgm^2)
N.C.	.251	.249	.169	2.008	1.147	.430	.0131	.0065	.0010
M.M.	.256	.258	.161	2.100	1.200	.450	.1142	.0073	.0010
R.S.	.275	.250	.160	1.966	1.123	.421	.0154	.0064	.0009
J.M.	.261	.256	.177	1.960	1.120	.420	.0138	.0067	.0011
D.H.	.274	.256	.171	2.156	1.232	.462	.0168	.0074	.0012
M.S.	.288	.275	.179	2.663	1.521	.571	.0229	.0106	.0016
D.S.	.292	.272	.175	2.346	1.341	.503	.0207	.0091	.0014
T.S.	.306	.307	.178	2.674	1.528	.573	.0260	.0132	.0016
P.O.	.274	.250	.178	2.472	1.412	.530	.0192	.0088	.0015
J.P.	.290	.272	.202	2.346	1.341	.505	.203	.0091	.0018

Table 6

Body Segment Parameters of Class II Wheelchair Basketball Players

	Arm Length (m)	Forearm Length (m)	Hand Length (m)	Arm Weight (kg)	Forearm Weight (kg)	Hand Weight (kg)	Arm Moment of Inertia (kgm²)	Forearm Moment of Inertia (kgm²)	Hand Moment of Inertia (Kgm²)
D.K.	.291	.270	.188	2.380	1.360	.510	.0209	.0091	.0016
S.R.	.251	.255	.172	1.700	0.971	.364	.0111	.0053	.0009
M.E.	.279	.255	.168	2.128	1.216	.456	.0171	.0072	.0011
M.K.	.268	.257	.172	2.436	1.392	.522	.0181	.0084	.0014
W.E.	.256	.274	.167	2.464	1.408	.528	.0167	.0097	.0013
M.V.	.250	.266	.156	2.324	1.32[illegible]	.498	.0151	.0086	.0010
S.D.	.293	.247	.171	2.173	1.242	.466	.193	.0082	.0012
A.L.	.259	.247	.150	2.033	1.162	.436	.140	.0065	.0008
R.P.	.275	.258	.190	2.184	1.248	.468	.0184	.0076	.0015
R.W.	.250	.270	.165	2.184	1.248	.468	.0142	.0084	.0011

Table 7

Body Segment Parameters of Class III Wheelchair Basketball Players

	Arm Length (m)	Forearm Length (m)	Hand Length (m)	Arm Weight (kg)	Forearm Weight (kg)	Hand Weight (kg)	Arm Moment of Inertia (kgm^2)	Forearm Moment of Inertia (kgm^2)	Hand Moment of Inertia (Kgm^2)
J.W.	.230	.196	.165	2.313	1.322	.496	.0127	.0047	.0012
A.C.	.275	.283	.170	2.520	1.440	.540	.0198	.0106	.0014
J.M.	.280	.271	.199	3.125	1.780	.670	.254	.0121	.0023
A.R.	.311	.268	.177	2.184	1.248	.468	.0218	.0082	.0013
B.D.	.288	.268	.170	3.298	1.885	.707	.0283	.0124	.0018
P.K.	.293	.273	.177	2.052	1.173	.440	.0182	.0080	.0012
T.B.	.288	.278	.166	2.604	1.488	.558	.0209	.0073	.0010
T.	.261	.264	.160	1.996	1.141	.428	.0141	.0073	.0010
C.B.	.317	.268	.202	2.755	1.574	.590	.0287	.104	.0021
R.S.	.270	.271	.160	3.290	1.880	.705	.0248	.0126	.0018

IX. APPENDIX D: Error Analysis

Many things may happen that can reduce the reliability and validity of results obtained in biomechanics research (Hudson, 1983). For example, when cinematography is used to obtain records of human motion, error can be introduced during data collection, data reduction and/or data analysis. Such error might be introduced by the measurement system, the subjects studied, the researcher and/or the environment.

While the noise (error) aspect of the obtained raw data may not be evident in displacement parameters of motion, the error becomes obvious at higher order derivatives (Patrick et al., 1980). Consequently, Cavanagh (1975) suggested that the area of numerical differentiation and the accompanying smoothing processes remained the weakest link in the chain of tools used for human movement analysis.

In this study, attempts were made to minimize controllable inaccuracies and to evaluate the accuracy and consistency of the obtained results.

1. A frame was randomly selected and then was digitized twice. A Pearson product-moment correlation was calculated to evaluate the consistency of the digitized body segments end points. A high positive correlation was found (r=0.999). This represents acceptable consistency during the data reduction process.
2. The calculated vertical acceleration of the airborne ball (after release), served as an indicator of the accuracy of the results obtained in this study. The calculated acceleration of the ball was the weakest link

in this study, since it not only included errors due to inaccurate estimation of the ball's CM, but the noise was magnified in the process of double differentiation. Consequently, an evaluation of the acceleration of the airborne ball against the known value of acceleration due to the force of gravity ($g=9.81$ m/sec^2) was thought to be the best indicator of accuracy for this study. It was found that the calculated acceleration of the ball after release fluctuated between -10.1 and -11.32 m/sec^2. The obtained result indicated taht the overall error introduced by all perspective errors, distortion errors, digitizing inaccuracies, and errors due to double differentiation, was smaller than 15 percent. It was speculated that the calculated angles and velocities were much more accurate since such data undergo at the most one mathematical differentiation.

3. A much less significant source of error was included in the calculation of body segment parameters. The estimated predicted body weight and segmental weight of class II wheelchair basketball players was the weakest link in this part of the present study, since it was difficult to estimate the weight reduced due to the physical disability. A subjective evaluation of the predicted body weight of the class II wheelchair basketball players in this study was conducted, and consquently, it was decided that 22 percent would be added to the actual body weight of class II wheelchair

basketball players in order to obtain their predicted body weight. Hypothetical examination showed that the error that might be introduced by inaccurate estimation of predicted body weight was relatively small. It was found that when the performance of class II wheelchair athletes in this study was repeatedly analyzed with three different hypothetical body weights (W1=predicted body weight; W2=predicted body weight plu[illegible] W3=predicted body weight minus 5 Kg), the [illegible]ed equivalent torques differed by no more than [illegible] Nm .