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Review

Walking assistance using crutches: A state of the art review

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ABSTRACT

Crutches are one of the most common ambulatory assistive devices. Using crutches encourages more physical activity than many other assistive devices, which has long-term health benefits. Recent advances have led to improvements in performance, but using crutches remains slower than normal walking, are energetically inefficient, cause additional strain on upper extremities, and often result in abrasions on the skin. Further improvements to address these deficiencies are needed but require an understanding of the crutch users' disabilities, different crutch gait patterns, associated biomechanics, and how the crutch design interacts with the user. It is important that research studies and designs take into account parameters from multiple ways of measuring performance in order for impaired users to achieve effective crutch walking. Many existing studies of crutches only analyze a subset of quantitative variables, so the overall impact of a design or modification is not fully assessed or comparable to other designs. Another important aspect is the user; each crutch type has specific characteristics that need to match the user's ability, physical fitness, and gait pattern. Pain and injuries on upper extremities should also be considered as an important factor in long-term users.

A search was done to find research papers and related patents focusing on crutch design and usage. Papers that studied one or more of the following topics were included: effects of crutches on the gait parameter, types of crutch walking patterns, improving walking efficiency through crutch design, and identifying the important components when studying a gait. This review paper summarizes the effects of existing crutch types and gives guidelines for how future studies should comprehensively evaluate design changes. This paper includes an overview of crutch gait walking patterns, users, the components and measurements of crutch studies, and advancements of crutch designs.

Contents

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1. Introduction

Crutches are used by many people for daily mobility because of impaired use of one or both legs. Around six million people in the USA use crutches, 4.2% of women and 3.4% of men in Europe have a walking disability, and around 2.8 million people in Canada have a type of mobility disability. These disabilities lead to 20 and 39 percent difference in employment rate between people with and without disabilities in Europe (Eurostat, 2011) and Canada (Morris et al., 2017), respectively. The number of individuals using assistive devices for mobility is growing more rapidly than the general population (Kaye et al., 2000; Russell et al., 1994). However, this reported number of crutch users is conservative since it does not include partially-impaired persons using a single or pair of crutches as a supplement to other assistive mobility devices such as wheelchairs, scooters, and lower limb prostheses (Dreeben, 2006). While a crutch user may opt out of crutch walking to predominately use a wheelchair, the use of crutches encourages upright posture, remaining active, and more independence, all of which are highly beneficial for long-term health (Deaver, 1933; Haubert et al., 2006).

Although crutches are beneficial for those that need them, studies have shown that crutch gait is slower (Mcbeath et al., 1974; Sankarankutty et al., 1979), has a higher energy cost (Mcbeath et al., 1974), and increases heart rate (Sankarankutty et al., 1979). It also changes joint displacement (Shoup et al., 1974), plantar foot pressure (Lee et al., 2011), and vertical and horizontal ground reaction forces (Stallard et al., 1980). Although many recent advances have improved crutch performance, further advancements are still needed.

This review paper presents a summary of crutch research and design over the past century and points out areas where additional research is needed. Section 2 presents an overview of crutch gait walking patterns, basic crutch structure, and crutch users. Section 3 details all the components and measurements of crutch studies including forces, energy consumption, and gait variables. Section 4 summarizes advanced crutch designs.

2. Crutch foundations

2.1. Crutch walking patterns

Although crutches allow ambulation for those with an impairment, walking with any type of crutch fundamentally changes the pattern of walking (Shoup et al., 1974). Smidt and Mommens (1980) showed that walking velocity in normal gait is faster than all other assisted gaits. Fig. 1 shows an example of the main differences between a normal gait and the swing-through crutch gait. Whereas the upper limbs are 180° out of phase in normal gait, they move in unison with crutches. The vertical fluctuation of the shoulders increases in this crutch gait but is relatively constant during normal gait. Another difference is that the trunk stays in a relatively upright position in normal gait while it is flexed in crutch strike. The flexed posture and oscillation of upper extremities alter the range of joint displacements and angles.

There are multiple ways to walk with crutches that depend on the specific injury or disability. The structure of crutch gaits vary based on the delay between the ambulatory device and foot placement, the number of concurrent points of contact, and laterality. A general convention for referring to ambulatory assisted gait (Smidt and Mommens, 1980) is defined in Fig. 2. Canes are included for comparison and completeness.

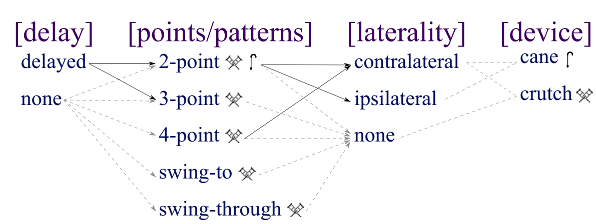

Fig. 2. A general method for naming ambulatory assisted gait. Arrows connecting each column indicate compatibility. For instance, a "delayed" gait is only compatible with 2-point and 3-point gaits.

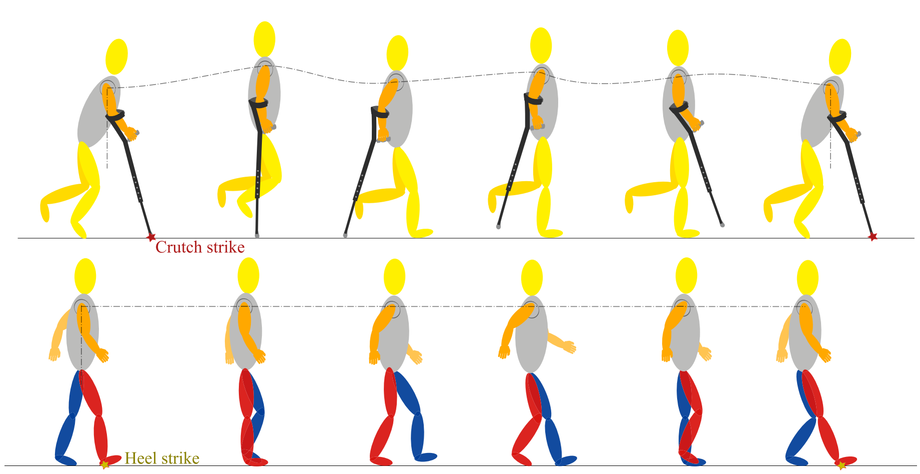

Fig. 1. Swing-through crutch and normal walking gait in one gait cycle. Comparing strike points, shoulder movements and trunk posture

Based on this convention (Smidt and Mommens, 1980), points or patterns indicate the basic structure of the gait. Points are the number of ground contacts in one gait cycle that leave the ground and land at the same time in a line perpendicular to the direction of walking. They can change from 2-point to 5-point depending on the device. Two other patterns, swing-to and swing-through, are when the lower body swings to or pass the device. Not all the points and patterns are compatible with an assistive device. Icons beside each point/pattern in Fig. 2 indicate device compatibility. For instance, a 3-point gait can only be used with crutches. Delayed is when the foot lands after the device and typically results in increased step length and time (Smidt and Mommens, 1980). Laterality refers to one device moving at a time. The number of counts is defined as the number of separate floor contacts that happen in one gait cycle. The number of counts in an assisted gait structurally alters the gait and affects gait variables such as cycle time, step time, stride length, double-stance time, and acceleration (Smidt and Mommens, 1980). Fig. 3 shows an example of normal, 2-point, and 4-point gait. The trajectory of the center of pressure (COP) significantly changes in 4-point crutch gait compared to normal. 2-point gait has a trend similar to normal walking and 4-point gait is significantly different from normal gait (Lee et al., 2011).

Partial Weight Bearing (PWB) indicates the percentage of body weight divided between a crutch and the lower limbs. A 10% PWB walking gait means that 10% of body weight is transferred to the impaired side while the crutch carries the rest. A non-weight bearing swing-through crutch gait has a single foot landing; meaning that no weight is transferred to the impaired leg (Fig. 4a) while a PWB swing-through gait lands on both feet (Fig. 4b) (Smidt and Mommens, 1980). Determining which gait is more beneficial depends on the lower-limb strength of the user and the impaired side. However, using a 3-point crutch gait (Fig. 4c) creates three ground contact points at the same time which allows for a range of PWB. A study comparing 10, 50, and 90% PWB (Li et al., 2001) indicated users could reproduce 50% PWB more accurately than the others.

Comparing 2-point, 3-point, and swing-through gaits with axillary and Lofstrand crutches, one study (Mcbeath et al., 1974) found that the energy cost of a 3-point non-WB gait is close to swing-through gait while a 3-point PWB is more similar to a 2-point crutch gait and a cane gait. The energy cost (oxygen volume per min) and efficiency (oxygen volume per meter) for 3-point non-WB and swing-through was higher than 3-point PWB and 2-point crutch gait.

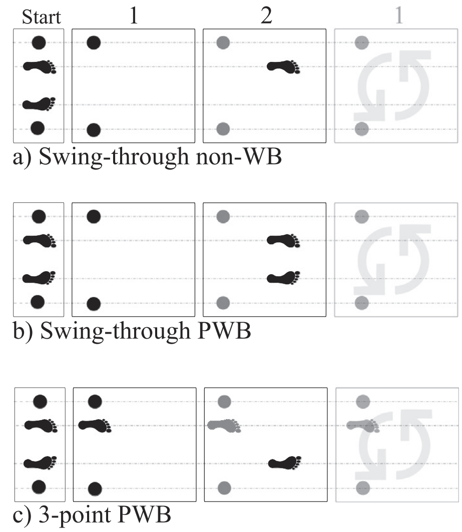

Fig. 4. Comparing weight bearing between 3-point and swing-through crutch gait.

2.2. Basic crutch designs

Fig. 5 shows some of the major crutch advances over the past century. Crutches are typically categorized as axillary and non-axillary designs (Lowman and Rusk, 1961; American Academy of Orthopaedic Surgeons, 1985). Forearm and axillary crutches are the most common types of crutches. The advanced designs will be presented in Section 4.

Axillary crutches were first used during the Greek era in Egypt (Loebl and Nunn, 1997) and generally include an underarm pad and have more trunk support than other crutch types (American Academy of Orthopaedic Surgeons, 1985). Weight bearing should only happen on the handle, and the axilla pad should only be used for stability. One of the first recorded patents, recorded in 1908 (Hargrove, 1908), introduced an axillary crutch with handles, armpit, and a telescope to adjust height.

Non-axillary crutches have many shapes and configurations. One of the most commonly known is the forearm crutch, where the underarm pad is replaced with an armband around the forearm to transfer body weight and reduce pressure from the underarm (Thorssen, 1940). They are suitable for patients with weak hip

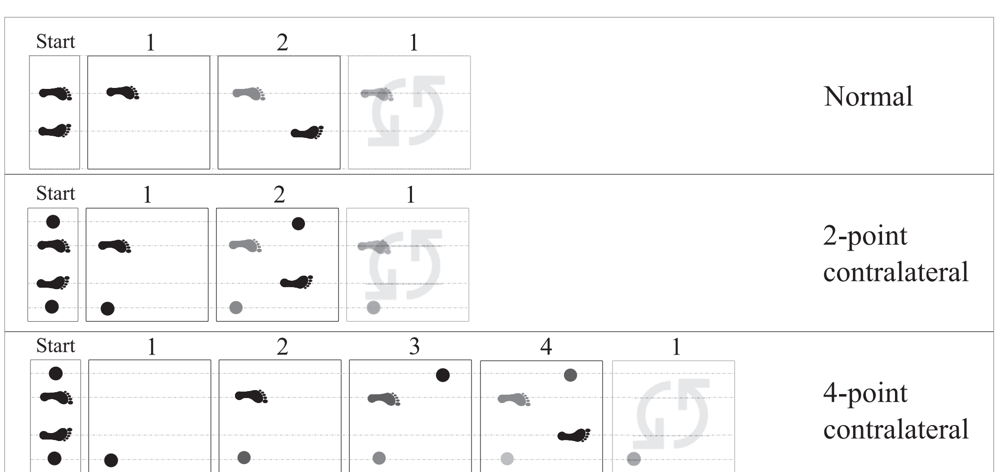

Fig. 3. Normal gait, 2-point, and 4-point crutch gait compared with the number of contact points and time frames. Each time frame indicate simultaneous ground contacts that land at the same time.

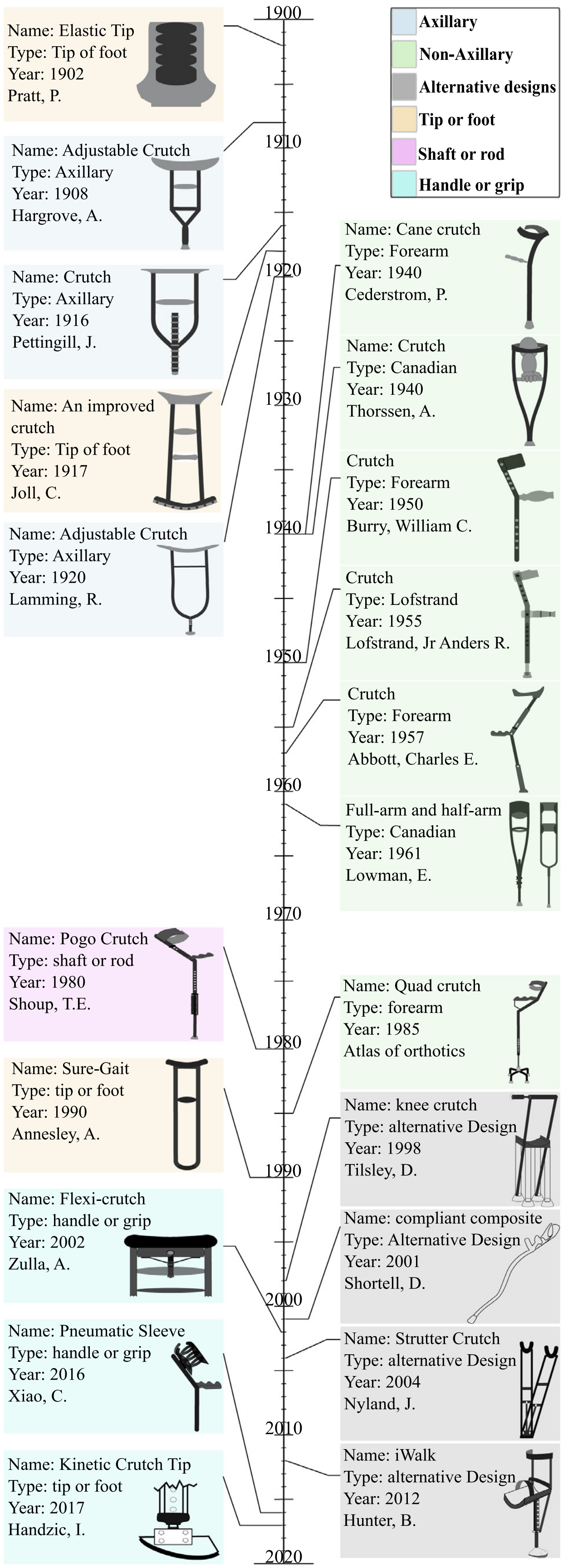

Fig. 5. Timeline of crutch evolution through design modifications. The designs on the left hand side focus on remodeling one part of the crutch (tips, shafts, or handles) while the right-hand side are designs including major alteration to the whole crutch structure.

abductors and extensors. They also have faster walking speeds compared to axillary crutches (American Academy of Orthopaedic Surgeons, 1985).

The Lofstrand crutch, introduced in 1948 (Lofstrand, 1948; Lofstrand, 1955), includes a handgrip and an attached cane (Fig. 5). The forearm clamp is at an angle so the arm can slip easier into the cuff. There have been other patented crutches with various modifications of the Lofsrand design (Burry and Christie, 1950; Abbott, 1957).

The Canadian crutch is another frequently used non-axillary crutch that combines features of both the forearm and axillary crutch (Stallard and Rose, 1978; Sankarankutty et al., 1979; Dounis et al., 1980). It creates more stability than a forearm crutch while having less underarm pressure than axillary crutches. A triceps Canadian has two cuffs around the forearm and above the elbow joint which could be full-arm or half-arm cuffs. Although half-arm gives better freedom of hand movements, full-arm can be more assistive in patients with less strength in their triceps (Lowman and Rusk, 1961).

2.3. Crutch users and experimental subjects

Crutches are necessary tools for a wide range of disabilities including partial paralysis, Spinal Cord Injury (SCI), Multiple Sclerosis (MS), etc. Each category requires assessment based on their own symptoms, limitations, and physical ability to prescribe the most efficient assistive technology. However, most experimental subjects used in research are healthy since they are more available and their assisted gait can be compared to their normal walking.

Able-body subjects cannot completely demonstrate the walking gait of a disabled crutch user when walking with a crutch due to differences in push-off ability, trunk range of motion, and full usage of shanks and knees (Noreau et al., 1995; Wells, 1979). Paraplegic patients have a lack of motor function in their gaits that affects their pattern of crutch gait (Noreau et al., 1995). Increasing restriction on lower extremities of able-body subjects indicates a more forward placement of crutches that causes the users to move from a swing-through gait to a swing-to gait (Wells, 1979). Comparing swing-through gait in paraplegic and able-body subjects indicated longer crutch stance phase in the paraplegic group due to slower hip flexion and greater physical strain on the upper body. Shorter stride length also contributed to changing the pattern of swing-through gait to swing-to in the paraplegic group (Noreau et al., 1995; Wells, 1979).

A study examining adults with SCI walking with forearm crutches indicated a significant difference between swing-through and 2-point gait in superior, posterior, and medial forces as well as joint trajectories in the shoulder (Perez-Rizo et al., 2017). The results of assisted walking between each category of disability should not be used interchangeably as research has shown patients with MS and SCI apply dissimilar propulsion forces when using ambulatory assisted technologies (Souza et al., 2010). Studies with incomplete SCI patients have shown that walkers significantly decrease speed and step length while increasing stability (Melis et al., 1999; Saensook et al., 2014). Therefore, it is vital to prescribe an assistive device that is appropriate to each individual.

Properly adhering to crutch gaits is also challenging for inexperienced users. Research (Goh et al., 1986) showed 34% of BW was carried under the arm when axillary crutches are used incorrectly. Current forearm designs have also been associated with hematoma formation and pain along ulna bone (Fischer et al., 2014). Incorrect adjustment of crutch height will also cause pain on the forearm and axillary muscles which can lead to nerve damage in long term.

3. Crutch study: components and measurements

Studying crutch gait requires understanding both human motions while locomoting and how the crutch designs interact with the person. Therefore, there are two major classes of features researchers should utilize to compare and analyze crutch gait. The first group of variables measures human performance, which are the physical or psychological responses of the human body during crutch walking. These variables include oxygen consumption (Ghosh et al., 1980; Thys et al., 1996), energy consumption (Wells, 1979; Sankarankutty et al., 1979; Ghosh et al., 1980; Thys et al., 1996), peak heart rate (Sankarankutty et al., 1979; Ghosh et al., 1980), maximum forces applied to the upper extremities (Wilson and Gilbert, 1982; Goh et al., 1986; Melis et al., 1999; Requejo et al., 2005), etc. This class also studies the long-term effects of incorrect crutch use, especially on upper extremities by measuring underarm, forearm, axilla, and hand pressure (Goh et al., 1986; Ginanneschi et al., 2009; Fischer et al., 2014; Requejo et al., 2005; Sala et al., 1998).

The second class incorporates dynamic aspects of crutch locomotion and includes kinematic measurements. These variables include velocity (Wells, 1979; Ghosh et al., 1980; Thys et al., 1996), displacement (Shoup et al., 1974; Wells, 1979; Noreau et al., 1995), and acceleration in the 3-dimensional planes (Tsuda et al., 2014; Dlugosz et al., 2017), ground reaction forces (Stallard and Rose, 1978; Stallard et al., 1980; Li et al., 2001; Wilson and Gilbert, 1982), gait phase (Lee et al., 2011; Goh et al., 1986; Smidt and Mommens, 1980; Noreau et al., 1995; Wells, 1979), joint angles (Shoup et al., 1974; Li et al., 2001; Melis et al., 1999), etc.

Variables in both classes are related, and the effect of both should be jointly studied to achieve a comprehensive analysis of crutch behavior. Table 1 summarizes the variables and their categories along with what has been evaluated in the literature. Note how many of the studies shown in Table 1 only compared a subset of the variables and tended to focus on variables in the same category. For instance, research that focused on angles did not study energy consumption. Most research that focuses on gait variables does not typically consider joint angles or crutch angles, and research concentrating on angles rarely study gait variables. However, it is important to know how variables in all categories are changing to thoroughly evaluate a crutch.

3.1. Forces and force measurement

Using crutches causes a significant change in the forces applied to the body. Whereas normal walking has forces only at the feet, crutch walking creates forces applied to the upper extremities, predominantly the hands and axilla in axillary crutches and the hands and forearms on the forearm crutches, as shown in Fig. 6. Crutches transfer ground reaction forces (GRF) to the upper limbs, which then carry body weight during the body swing phase of crutch walking gait.

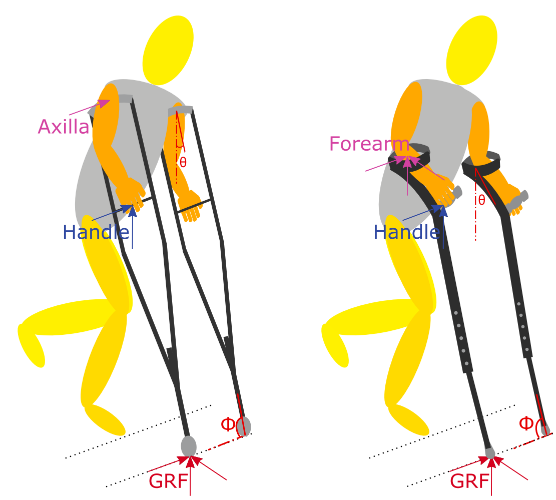

Fig. 6. Ground reaction forces and forces applied to upper extremities for both axillary and non-axillary (forearm) crutch. Crutch angles relative to the body and the ground. θ is the angle between a crutch and the vertical axis in sagittal plane and ϕ is crutch angle with the ground on the frontal plane.

Table 1
Main categories of variables in crutch gait studies. Solid dots indicate variables measured in the study, X labels are parameters calculated and compared, and S labels were set at a fixed value.

		Dynamics													Kinematics																	
		Forces						Energy Consumption							Kinematics and Gait variables																	
		GRF (Axial, A/P, M/L)	Forces on Upper Extremities	Moment of Forces	Center of Pressure Trajectory	Center of Gravity	Plantar Foot Pressure	Oxygen Consumption	Lactic Acid	Heart Rate	Kinetic Energy	Potential Energy	Energy Cost	Mechanical Power	Shoulder Angle	Trunk Angle	Pelvis Angel	Hip Angel	Knee Angle	Ankle Angle	Toe/Foot Angle	Range of Motion	Crutch Angles	Joint Displacement	Step\Stride Length	Cycle\Step Time	Double-Stance Time	Contact Time	Phase Ratio (swing/stance)	Cadence	Velocity\Speed	Acceleration
1974	Shoup														•			•	•	•				X								
1999	Melis		•													•		•	•	•					•					•	•	
2001	Sheng Li	•				X										•	•	•	•	•	•	X									•	
2009	Gil-Agudo																•	•		•					•					•	•	
2005	Requejo	•	X												•	•									•					•	•	
1982	Wilson	•	X																				•									
1978	Stallard	•																														
1980	Stallard	•																											•			
1980	Smidt																								•	•	•		•		•	•
1986	Goh		•																										•		•	
2011	Lee				•		•																					•	•			
2014	Tsuda																							•								•/X
2017	Dlugosz	•																							•					•	•	X
1979	Sankarankitty									•			X																		•	
1996	Thys							•	•		•	•	X	X																	S	
1980	Ghosh							•		•			X																		S	
1979	Wells										•	•	X	X										•	•	•			•		•	
1995	Noreau			•										•										•	•				•	•	•	•

Forces on the handle, axilla, and/or forearm can either be measured directly (Sesar et al., 2019; Lancini et al., 2016; Seylan and Saranlı, 2018; Mekki et al., 2017; Sardini et al., 2014; Tsuda et al., 2010; Chen et al., 2018; Varoto et al., 2014; Fischer et al., 2014) or derived indirectly based on the GRF and the crutch angles (Fig. 6) (Wilson and Gilbert, 1982). The direct approach typically uses motion capture systems, pressure mats, and/or wearable sensors. Wearable sensors are more accessible and can be used outside of the lab environment (Lancini et al., 2018). An instrumented crutch is equipped with sensors that measure GRF and accelerometers to measure angles because the forces applied to a crutch are tilted. Researchers have developed algorithms that accurately estimate crutch pitch angles (Sesar et al., 2019). Machine learning models have been used to calibrate the accuracy of instrumented crutch results (Chen et al., 2018). A low-cost forearm crutch was designed using quadratic grid arrangement of force sensitive resistors to estimate the GRF and angles (Seylan and Saranlı, 2018). A wireless connection is important for instrumented walking assistance devices (Mekki et al., 2017; Sardini et al., 2014). Mekki et al. (2017) introduced a wireless cane that can measure gait parameters in Parkinson's patients and predict gait dysfunctionality. Another study evaluated an instrumented crutch on ten healthy subjects and indicated an accurate estimation of crutch angle and angular velocity (Tsuda et al., 2010).

Wilson and Gilbert (1982) concluded that approximately 7.5% of body weight is applied to the axilla normal to the sagittal plane at the apex of swing-through gait. They also found that a total average of 1.8 times of body weight is applied to the hands through the crutch handles. Another study has also found that between 111% and 120% of BW can be applied to upper extremities during walking and running with Lofstrand crutch (Tatar et al., 2018). Goh et al. (1986) found that the posterior force applied to hands is compressive when crutches are used incorrectly and tensile when used properly. Measuring pressure distribution on the forearm when using a forearm crutch has indicated that the highest maximum and average pressures were on intermediate and ulnar quadrants and increasing the load on the crutch creates a horizontal shift of COP from intermediate to the ulnar region (Fischer et al., 2014).

Researchers have derived the relationship between motions, forces, and momentums of the upper limbs and joints in order to make a connection between upper limb injury and crutch gait in long-term crutch users (Slavens et al., 2010; Slavens et al., 2011; Perez-Rizo et al., 2017; Requejo et al., 2005). Deriving inverse dynamics of upper limbs indicated larger peak forces and bigger motions at joints in swing-through gait compared to 2-point gait (Slavens et al., 2010). Applying these formulations to patients with myelomeningocele (Slavens et al., 2010), cerebral palsy, SCI, and osteogenesis imperfecta (Slavens et al., 2011) indicated that the greatest peak forces when using forearm crutch were in the sagittal plane while the greatest moments were in the shoulder. Comparing the same two gaits in patients with SCI using a forearm crutch indicated significantly higher peak forces in all directions as well as flexo-extension torque with swing-through gait (Perez-Rizo et al., 2017). Research with a Lofstrand crutch using a 4-point gait with an incomplete SCI patient indicated greatest peak net force and moments were in the upper extremity opposite of weaker foot (Requejo et al., 2005). Understanding the inverse dynamics can help optimize rehabilitation methods and prescribe the most efficient gait for individual conditions.

Walking on crutches increases the GRF applied to lower extremities. Stallard and Rose (1978) showed that the maximum GRF in axillary crutch walking exceeded body weight by 3–18% compared to normal walking. Using a single foot in crutch gait creates about 25% higher peak force than normal gait and about 33% higher when both feet landed together. Canadian crutches follow a similar pattern (Stallard et al., 1980), but the maximum vertical forces and the peak horizontal forces increased. Another study (Goh et al., 1986) concluded that crutch walking increases peak vertical forces by 21%. The authors also indicated that 44% of body weight is applied to upper extremities in crutch gait and 34% of body weight is applied to the underarm when the incorrect crutch gait is used.

One of the less studied features in crutch research is plantar foot pressure. Lee et al. (2011) found significantly less pressure on the forefoot in 4-point crutch gait compared to 2-point crutch and normal gait. Crutch users rely on the crutch and less on their foot to keep their balance, which creates forward propulsion in the body and causes a flexed trunk. The authors concluded that increasing contact time and using upper extremities for push up instead of ground reactions could be possible reasons for this shift in the COP.

3.2. Energy consumption

Energy consumption is often measured through indirect methods such as measuring oxygen consumption (Thys et al., 1996; Ghosh et al., 1980), heart rate (Sankarankutty et al., 1979; Ghosh et al., 1980), or potential and kinetic energy (Wells, 1979; Thys et al., 1996). If efficiency remains constant, energy consumption is linearly correlated to mechanical work (Thys et al., 1996). Fig. 7 shows their respective components. Energy consumption consists of oxygen consumption and anaerobic glycolysis; however, the contribution of anaerobic glycolysis is between 1–8% and can typically be neglected (Thys et al., 1996). Total mechanical work contains both external movements of the body and the internal movement of the limbs.

Comparing energy consumption and mechanical work in crutch walking gait (Thys et al., 1996) indicated approximately 50% efficiency reduction compared to normal gait. Therefore, although energy consumption has increased by a factor of 2.5 times in crutch gait, the applied muscular work has increased only 1.4 times. There are several reasons for the decrease in efficiency. First, crutch walking is less stable, which consequently results in more energy expenditure. Secondly, crutch movement requires the use of upper and lower limbs versus only lower limbs in normal walking. Thirdly, crutches are more rigid, and there is a delay between feet and crutch landings compared to normal walking.

Heart rate and oxygen consumption are two measuring approaches for estimating energy cost (Sankarankutty et al., 1979). Both methods are comparable for measuring energy if other conditions stay the same (Ghosh et al., 1980). Walking speed affects energy cost and efficiency. Energy expenditure of crutch walking is significantly higher than normal walking at a slow speed, but they are closer when walking faster (Ghosh et al., 1980).

Axillary crutches require more energy consumption than Lofstrand when 2-point or 3-point non-weight bearing gait is used (Mcbeath et al., 1974). Energy consumption for axillary crutches increases with speed but the curve plateaus for forearm crutches. Canadian crutches have the lowest energy consumption (Stallard and Rose, 1978; Sankarankutty et al., 1979). Fig. 8 shows a summary of changes between these crutches by comparing heart rate and oxygen consumption. While comparing heart rate did not indicate a significant difference, comparing oxygen consumption did (Fig. 8b) (Dounis et al., 1980).

3.3. Gait variables

A comprehensive study of crutch gait requires kinematic data related to gait variables. Gait variables include angles and joint displacement, cycle time, step time, double-stance time, contact time, phase ratio, step length, stride length, cadence, velocity, and acceleration. Velocity is the most calculated variable and is either measured (Melis et al., 1999; Li et al., 2001; Sankarankutty et al., 1979; Wells, 1979; Noreau et al., 1995; Smidt and Mommens, 1980; Goh

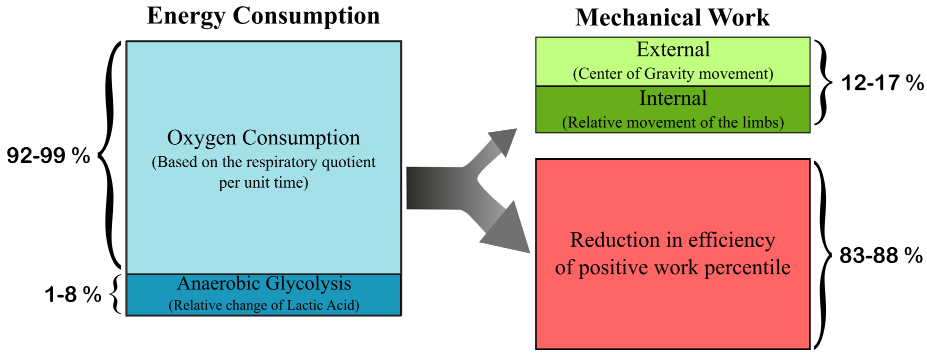

Fig. 7. Energy consumption and mechanical work elements in a crutch gait and their relation with efficiency of positive work percentile. Figure not drawn to scale for clarity in diagram.

Crutch Type	% increase $\frac{\textit{Heart Rate}}{\textit{Resting Heart Rate}}$	% decrease $\frac{\textit{Crutch Walking Speed}}{\textit{Normal Walking Speed}}$
Axillary	62 - 122	14 - 44
Elbow	44 - 101	15 - 45
Canadian	28 - 80	9 - 38

(a)

Oxygen uptake:	Canadian < Axillary < Elbow (F)
Distance walked in 6 minutes:	Canadian > Axillary > Elbow (M)
Distance walked in 1 liter of oxygen uptake:	Canadian > Axillary > Elbow (M)
Effort scale:	Canadian < Axillary < Elbow

(b)

Fig. 8. comparing Canadian, elbow and axillary crutch. (a) relative heart rate and walking speed percentile (b) energy cost and efficiency indicators. Solid lines show significant difference between all groups (both male and female) and dashed lines are only significantly different between one group of subjects (M: male or F: female).

et al., 1986; Dlugosz et al., 2017) or set to a desired value (Thys et al., 1996; Ghosh et al., 1980).

3.3.1. Angles and joint displacement

Joint angles are useful for comparing the posture in crutch walking. Joint displacement and deviation from the average value in the sagittal and frontal planes can be derived by measuring angles in the shoulder, hip, knee, and ankle (Shoup et al., 1974). The trunk and pelvis joints lean forward in crutch walking, and their range of motion decreases (Li et al., 2001). In normal walking, joints further from ground contact, such as the shoulder, have less variation (Shoup et al., 1974). However, in crutch gait, the ground contact point is shifted to the handle, which causes the shoulder to have the most variation (Shoup et al., 1974). Increased internal and external rotation on the hip cause a shift in the center of gravity in crutch gait (Li et al., 2001).

Wells (1979) found that increasing restrictions on joint angles decreased the percentage of body swing phase of crutch gait. However, Noreau et al. (1995) found that paraplegic patients had longer body swing phase due to slower hip flexion when using angular displacements of the hip, shoulder, and elbow. A study among patients with Central Cord Syndrome (CCS) indicated decreased range of motion in the ankle when using two crutches instead of one (Gil-Agudo et al., 2009). This can result in reduced risk of falling due to foot drop.

3.3.2. Gait time, length, and phase ratio

Measurements of time, such as contact time, can indicate a gait pattern change. One study (Lee et al., 2011) found a significant increase in mid-stance and decrease in heel contact time for crutch gait. The authors suggested that changes in stance phase would affect the stability and COP as a result.

Stride length is a gait variable that has been used as a measurement for the amount of a disability (Wells, 1979; Noreau et al., 1995). Increased disability in paraplegic individuals has been associated with significantly shorter stride length (Noreau et al., 1995). Cadence is related to the inverse of stride length; meaning increasing cadence is accompanied by decreased stride length (Noreau et al., 1995).

Phase ratio or swing/stance ratio compares the relative percentage of each stance or swing portion during one step. The portion of stance and swing phase in crutch gait compared to normal walking changed from 50% on each leg to approximately 44% on crutches (Stallard et al., 1980). Double-support phase increased when increasing disablement from no restrictions to full brace (Wells, 1979). Users need more time to swing crutches, which results in single leg stance phase increasing by 11% in crutch gait (Goh et al., 1986). This study also indicated a decrease in walking speed in crutch gait, which is consistent with other research (Lee et al., 2011). Based on another study (Noreau et al., 1995), healthy and paraplegic subjects both had the same walking speed preference; however, swing and stance phases differed between them. Subjects with a disability had longer crutch stance phase due to the lack of activation in the hip and slower hip flexion. The authors concluded that this would cause higher shoulder and elbow acceleration/deceleration and consequently greater physical strain on the upper extremities.

Due to the fundamental difference in a crutch and normal gait, a 3-point gait has a shorter stance phase and longer swing phase on crutches. Consequently, changes in walking velocity and hip rotation caused the center of gravity to shift toward the healthy leg (Li et al., 2001).

3.3.3. Velocity and cadence

Velocity is the most common variable in crutch studies. Most experiments measure velocity as a variable in the test, but some experiments set it at a fixed value (Thys et al., 1996; Ghosh et al., 1980; Wells, 1979). Self-selected speed for all ambulatory assisted gaits is significantly less than normal gait (Mcbeath et al., 1974). More importantly, self-selected speeds in ambulatory assisted gaits (between 50 and 65 m/min) are less than the most efficient speed range for these gaits (around 70 m/min) (Mcbeath et al., 1974). The result from multiple studies indicated a decreased velocity and cadence in all crutch gaits compared to normal gait (Sankarankutty et al., 1979; Noreau et al., 1995).

3.3.4. Acceleration

Stable walking has periodic acceleration while an unstable gait will have aperiodic acceleration (Tsuda et al., 2014). This relationship makes acceleration an important variable to measure. It can be measured directly using an accelerometer (Dlugosz et al., 2017) or through indirect methods. One example (Tsuda et al., 2014) aims to calculate user acceleration by running a linear regression between the acceleration ratio and body movement ratios. They concluded that there is a linear correlation between thigh movement ratio and acceleration. However, they only recorded four steps in each experiment which might not be enough to extend these results to all crutch walking gaits (Tsuda et al., 2014). Acceleration strongly depends on gait counts (Smidt and Mommens, 1980).

3.4. Measurement conclusion

When conducting crutch studies, parameters from all categories should be measured and compared for a comprehensive analysis. However, Table 1 indicates that most studies only focus on one area. Studies concentrated on energy consumption do not measure forces or joint angles and vice versa, but both categories are important.

Forces are critical parameters in comparing normal and crutch gait. Instrumented crutches have been used to provide biofeedback of forces for learning and training purposes. Research has shown larger forces applied to both upper and lower extremities during crutch gait, which is in agreement with higher energy consumption. Energy cost can reasonably be obtained through indirect methods. Oxygen consumption and heart rate are commonly used as an indicator of energy cost. Mechanical work is also linearly correlated to energy if efficiency does not change.

Gait variables range from joint angles to velocity and acceleration. Change of ground contact in crutch gait maximizes the displacement on shoulders relative to other joints. Velocity significantly decreases in crutch gait relative to normal. Some research also set velocity at a fixed value when using a treadmill for experiments to compare changes of other variables without the effect of velocity. Research has indicated self-selected speed in normal walking is close to optimal walking speed; however, crutch users tend to walk slower than optimal crutch speed. It can be concluded that crutch users require more training for walking at faster speeds.

It is important to note that different disabilities affect gait parameters differently. An improvement in the results of an assisted gait within one pathology cannot necessarily be extended to another. Table 2 summarizes all the critical aspects of crutch experimental research including pathology, devices, gaits, and measured parameters during past forty years. As an example of differences between pathologies, Souza et al. (2010) found while SCI patients benefited from a manual wheelchair, MS patients did not benefit because some of their applied forces to the wheelchair opposed forward propulsion which resulted in wasted energy. Therefore, while suggesting a device or gait for one disability can improve performance, the same cannot be said for another.

4. Crutch design modifications

An important aspect of studying an assisted gait is the ambulatory assistive device. Many recent studies are more focused on one aspect of performance of a crutch rather than looking at the full picture of crutch-user-environment. There are many factors that

Table 2
Main aspects of crutch studies in the past four decades including: pathology, device, gait, and measured parameters. Acronym used in the table are Knee Ankle Foot Orthosis (KAFO), Spinal cord injury (SCI), Post-polio syndrome (PPS), Central cord syndrome (CCS), Cerebral palsy (CP), and Osteogenesis imperfecta (OI)

Paper	Subjects	Devices	Crutch Gaits	Parameters
Smidth 1980	Healthy	Cane	2-point	Kinematics: speed, step time, step length, cycle time, swing/stance ratio, step time, double stance time, acceleration
		Crutch (N/A)	2-point, 3-point, 4-point	
		Walker	Delayed 5-point	
Sankarankitty 1979	Healthy	Crutch (Axillary, Elbow, Canadian)	Swing-through	Kinematics: speed Energy: heart rate
Lee2011	Healthy	Crutch (Axillary)	2-point, 4-point	Kinematics: contact timephase ratio Forces: plantar foot pressure
Li 2001	Healthy	Crutches (Axillary)	3-point	Kinematics: trunk, Pelvis, hip, knee, ankle, and feet angles Forces: GRF
Tsuda 2014	Healthy	Crutch (Axillary)	3-point	Kinematics: joint displacement, acceleration
Wilson 1982	Healthy	Crutch (Axillary)	Swing-through	Kinematics: crutch angles Forces: GRF, forces on handle and axilla
Goh 1986	Healthy	Crutch (Axillary)	Swing-through	Kinematics: speed, phase ratio Forces: forces on handle
Dounis 1980	Healthy	Crutch (Axillary, Elbow, Canadian)	N/A	Kinematics: distance Energy: oxygen consuption
Mullis 2000	Healthy	Crutch (Forearm)	3-point	Kinematics: speed Energy: oxygen consumption, heart rate
Nielson 1990	Healthy	Crutch (Axillary)	Swing-through, 2-point non-weight bearing	Kinematics: speed, stride length Energy: heart rate, oxygen consumptiom, energy cost
Parziale 1989	Healthy	Crutch (Axillary)	Swing-through	Kinematics: speed Forces: forces on handle
Segura 2007	Healthy	Crutch (Axillary)	Swing-through	Kinematics: stance time, stride length, speed Forces: GRF
Zhang 2013	Healthy	Crutch (Axillary)	Swing-through	Kinematics: stride length Energy: energy cost, mechanical power
Bertolaccini 2017	Healthy	Crutch (Axillary)	N/A	muscle activity
Capecci 2015	Healthy	Crutch (Axillary)	Swing-through	Kinematics: step length and time Forces: GRF Energy: heart rate
Rasouli 2017	Healthy	Crutch (Axillary)	Swing-through	Kinematics: speed, step time, step length, crutch ROM Forces: GRF
Annesley 1990	Healthy	Crutch (Axillary)	Swing-to	Energy: heart rate, oxygen consumption
Dooley 2015	Healthy	Crutch(Forearm)	N/A	Kinematics: cadence, stride time, foot contact, stride length, speed Forces: GRF
MacGillivray 2016	Healthy	Crutch(Forearm)	Swing-through	Kinematics: speed, phase ratio Forces: GRF
Sesar 2019	Healthy	Crutch (Axillary)	N/A	Kinematics: crutch angles Forces: GRF
Tsuda 2010	Healthy	Crutch (Axillary)	N/A	Kinematics: crutch angles and angular velocity Forces: GRF
Stallard 1980	Healthy	Crutches (Canadian)	Swing-through	Forces: GRF
Varoto 2014	Healthy	Crutch (Lofstrand)	2-point contralateral	Forces: GRF
Fischer 2014	Healthy	Crutch (Elbow)	3-point	Forces: presssure on upper extremities, Center of Pressure
Shoup 1980	Healthy children	Crutch (Forearm)	Swing-through	Kinematics: stride length, joint displacement Force: forces on handle
Sala 1998	Healthy & long-term crutch users	Crutch (Forearm)	3-point	Forces: Pressure on upper extremities
Wells 1979	Healthy with lower limb brace	Crutch (Axillary)	Swing-through	Kinematics: speed, support/swing phase Energy: potential and kinetic
Thys 1996	Healthy & temporary injured patients	Crutch (Elbow)	Swing-through	Energy: lactic acid, energy cost, oxygen consuption, kinetic energy, mechanical power
Ghosh 1980	Healthy & handicapped	Crutches (Axillary)	N/A	Energy: heart rate, energy cost, oxygen consuption
Noreau 1995	Healthy & Paraplegic	Crutch (Forearm)	Swing-through, Swing-to	Kinematics: speed, cadence, stride length, stance/swing percentage, acceleration Energy: Mechanical power
Tatar 2018	Amputee football players	Crutch (Lofstrand)	Swing-through	Forces: forces on upper and lower extremities
Leblance 1993	Healthy with KAfO, SCI, amputee, & PPS	Crutch (Axillary)	N/A	Kinematics: speed Energy: heart rate
Gil-Agudo 2009	CCS	Crutch(Forearm)	2-point	Kinematics: speed, stride length, cadence, phase ratio, ankle ROM
Slaven 2011	CP, SCI, OI	Crutch (Forearm)	2-point	Kinematics: joints ROM Forces: forces and torques on upper limbs
Freddolini 2018	post total hip replacement	Crutch (Forearm)	N/A	Kinematics: speed, cadence, stride length and time, shoulder ROM Forces: GRF, forces on upper limbs
Perez-Rizo 2017	SCI	Crutch (Forearm)	Swing-through, 2-point reciporocal	Kinematics: cycle length, stance/swing percentage, speed, cadence, stride length, ROM, joint trajectories Forces: forces and torques at shoulder
Saensook 2014	SCI	Cane	N/A	Kinematics: speed and distance
		Crutch (Forearm, Axillary)	N/A	
		Walker	N/A	
Melis 1999	SCI	Cane	N/A	Kinematic: speed, cadence, step length Forces: medial/lateral,
		Crutch (Elbow)	4-point	
		Walker	N/A	
Requejo 2005	Incomplete SCI	Crutch (Forearm)	4-point	Kinematics: speed, cadence, stride length, upper limb joints' angles Forces: GRF, Forces on upper extremities

should be taken into account when creating a new design. Each device has its specific characteristic that should be considered when being prescribed. As an example, Melis et al. (1999) showed that the percentages of supported body weight range from 100% in walkers to less than 50% in crutches and less than 25% in canes. While both crutches and canes create a restriction in movement and propulsion, walkers predominantly provide restrictions normal to the direction of walking. Walkers have the smallest step length and cadence; therefore, they have the slowest gait among these three (Melis et al., 1999). Canes have the fastest gaits because of the longer step length and higher cadence. Cane users had the most upright trunk posture and walkers created the most flexed trunk (Melis et al., 1999).

Research (Melis et al., 1999) has indicated pain and injuries on upper extremities should be considered as an important factor in long-term users in all three devices, so upper body fitness of users should be taken into account when prescribing an assistive device. Choosing an appropriate device should include the positive and negative side effects of each device, gait-training strategies to avoid injuries in long-term, and matching user's needs with physical capabilities. To address these user needs, many modifications have focused on modifying common crutches. These designs can be categorized into three different parts of a crutch:

I. Handle (upper side of the crutch including cuffs, forearm, and underarm pads): This section focuses on optimizing the weight bearing on upper extremities by suggesting new designs for elbow cuff, fixing wrist position, and repositioning handle and underarm pads. The aim of these designs is to reduce the risk of neurovascular damage to upper limbs.
II. Shaft (the middle section connecting the upper side to the tip): The majority of modifications in this section contain shock absorbers to reduce ground impact force to upper extremities.
III. Tip (lower section including the foot of the crutch): Designs involving the tip of the crutch frequently deal with the stability of walking and GRF. There have been fewer modifications in the tips than shafts and handles.

4.1. Handle or grip

Chronic crutch users often sustain injuries on their upper extremities because crutch gait partially transfers body weight to the upper limbs. Long-term use of crutches generates higher pressures on the palms, wrists, forearm, axilla, and shoulders for extended periods. These forces applied to the handle can indicate improper crutch use (Goh et al., 1986). Research has indicated case reports of ulnar tunnel syndrome after even short-term use of bilateral forearm crutches (Ginanneschi et al., 2009). The forces applied to the handle, axilla, and/or forearm are discussed further in Section 3.1. Designing proper handles can reduce the risk of injury.

The design of the handles in crutches can affect weight bearing on upper limbs. A handle in a forearm crutch with an extended elbow has less forces and momentum on the shoulder compared to a flexed elbow design (Freddolini et al., 2018). Research (Bertolaccini et al., 2017) has also shown that smaller handles with 20 mm diameters increase flexor digitorum muscle activity as detected by surface electromyography (sEMG) when compared to 40 mm handles. Higher activity in this muscle was also accompanied by higher perceived exertion on users. Comparing cylindrical and wide shaped handles in forearm crutches indicated a similar pattern. Maximum vertical forces was found in distal radial and middle palm for cylindrical and palm's proximal ulnar region for wide handle. However, researchers concluded they both showed similar pressure distribution pattern because 60–70% of average palmar loads was in radial side for both and therefore one can not be recommended over the other (Sala et al., 1998).

One of the inventions attempting to reduce extra loading on upper extremities is a pneumatic sleeve for Lofstrand crutches (Xiao et al., 2016). This research suggested a sleeve to fixate the posture of the wrists and decrease pressure on them by redirecting forces to cuffs (Fig. 5). This design includes a modified actuator in helical form and two half-cylindrical splints. Using the pressure applied to the crutch tip, the length of the helical actuator decreases, hence holding the wrists in place and preventing distal movements. Removing the tip depressurizes the actuator. A design for axillary crutches (Zulla and Colardo, 2002) created a space in between the underarm pads and the rest of the crutch by adding a spring in between (Fig. 5). This patent aims to decrease the risk of pressure on axilla by absorbing the shock through the spring and generates more flexibility with a pivoting handle.

4.2. Shaft or rod

Researchers have developed various methods for adjusting the length of the crutch based on the subject's height, length of the axillary fold to the heel, or arm span (Bauer et al., 1991). Although the methods differ, crutch height can vary by 2.5 cm with no significant difference in energy cost (Mullis and Dent, 2000).

Shock absorption is most frequently added to the shaft by adding springs or dampers, but other methods create an S-shape compliant shaft (Shortell et al., 2001). Measuring shock wave amplitude at the time of the crutch strike indicated a decrease of 22% in the spring-loaded axillary crutch (Pariziale and Daniels, 1989). Maximum forces on wrists and hands were also reduced by 24%. Seemingly contrary research (Segura and Piazza, 2007) has found an increase in the maximum GRF when using spring-loaded crutches. While the large GRF is measured at the tip below the spring, a reduced force to the hand and wrist is measured at the handle above the spring. Thus, the decrease in shock wave and maximum forces on hands and wrists indicates that the energy storage capacity of spring-loaded crutches can result in less fatigue and pain on upper extremities. Shortell et al. (2001) provided a method based on body weight to choose an appropriate stiffness for the desired result.

Additionally, reduced rate of GRF could help reduce the abrupt change of forces and the consequent shocks applied to upper limbs. Comparing spatiotemporal parameters (Segura and Piazza, 2007) indicated lower velocity, higher stride time, and higher crutch stance time in spring-loaded axillary crutches; although, stride length remained the same. One explanation is that the flexible structure of spring-loaded axillary crutch makes this design more difficult to control. Therefore, gaining balance comes at the expense of making users feet land closer to the crutch (Segura and Piazza, 2007; Zhang et al., 2013).

Although metabolic cost and mechanical cost decreased in spring-loaded axillary crutches, the ratio of mechanical cost to metabolic cost showed a significant increase in the spring-loaded axillary setup (Zhang et al., 2013). This increase in mechanical efficiency can be related to the additional energy storage capacity during the crutch stance phase.

Springs have also been incorporated into non-axillary crutches (Olivera, 2001; Shoup, 1980). One design, the pogo crutch (Shoup, 1980), integrates a telescopic spring into a forearm crutch (Fig. 5). The design reduces the center of mass fluctuation and stores impact forces at crutch strike as mechanical energy used for toe-off. Experiments showed changes in vertical forces, impact forces at initial contact were reduced by 50%, and toe-off peaks were eliminated. This result demonstrates that mechanical energy stored in springs could assist with the push-off. However, there is a tradeoff between energy storage capacity and stability.

Other designs have used an elastomeric system (damper) to eliminate the sense of instability due to height changes in springs. They found no significant difference in GRF, spatiotemporal parameters (Dooley et al., 2015), and peak vertical forces (MacGillivray et al., 2016). However, a mechanical test indicated a reduction in peak GRF when a large force (greater than the human range) is applied (Dooley et al., 2015). Also, experiments within a male group showed larger propulsive and smaller braking forces. This can be an indication of potential changes in the forward momentum of walking (MacGillivray et al., 2016).

4.3. Tip or foot

The invention of the crutch tip goes back to the 18th century (Tuttle, 1885). The first one was a tip with a screw-on rubber foot with a metallic thread attached to the crutch bar and an elastic cover at the foot. Crutch tip designs have seen less development compared to the handle or shaft. The most common is a circular rubber tip with a flat bottom.

Most of the advances in crutch tips have focused on improving the stability, balance, and traction. The first patent (Philip et al., 1902) details an elastic socket with a ring-shaped flange made of vulcanized rubber (Fig. 5). Other patented designs have added rubber cushions to prevent slipping (Pratt, 1910), a suction grip to compress air for a better hold on the ground (Candido, 1959), and inserting rigid materials to transfer forces to the sides (Urban, 1973).

Most of the patents have incorporated the same idea of a rubber tip with a flat surface; however, few designs have included a curved tip. The roller crutch (Joll, 1917) was similar to an axillary crutch at the top, but the two vertical bars diverged at the bottom and were connected to a curved hardwood with an attached rubber called "the roller" (Fig. 5). Experimental testing of the rolling design reported improved stability; 16% increase in step length and an increase in the effective supporting angle by 16 degrees. However, only one subject was tested for these results. The rolling crutch did not see any improvements for 80 years until the Sure-Gait and Rocker-bottom axillary crutches were evaluated, but there were no differences between the Sure-Gait (Fig. 5) and standard axillary crutches (Annesl et al., 1990; Nielsen et al., 1990). Although both crutches required significantly more energy expenditure than normal gait, they had similar oxygen uptake and heart rate (Annesl et al., 1990; Nielsen et al., 1990).

Since the crutch tip is the ground contact point, it can play a critical rule in improving the dynamics of crutch users by redirecting applied forces. A non-constant radius tip named the Kinetic Crutch Tip (KCT) shifts the ground contact point backward, which generates a moment (Handzic and Reed, 2017). This moment creates a controlled imbalance by redirecting vertical forces into a horizontal motion (Fig. 5). Depending on the orientation, the KCT can generate an assistive force (for flat/uphill walking) or a resistive force (for downhill walking) (Capecci et al., 2015). Horizontal forces parallel to the movement are assistive and the ratio of assistive to resistive is an indicator of crutch performance. Symmetric tips show an equal amount of assistance and resistance as the tip rotates from negative or positive. Because the KCT is asymmetric, it can generate more assistive forces and less resistive forces throughout the range of motion that help propel the user forward (Rasouli et al., 2017).

4.4. Alternative crutch designs

The majority of the presented research has only applied modifications to common crutch types in order to create more comfort or improve the performance. Only a few attempts have redesigned the whole crutch structure.

One design (Shortell et al., 2001) used a one-piece composite compliant crutch that incorporated an angled-cuff similar to a forearm crutch, wrist support, and a curved 'S' shaped main body to absorb shock (Fig. 5). Although the analytical experiments indicate reduced energy consumption using stored energy during the gait cycle, user feedback indicated instability and discomfort due to the compliant structure.

Another design (Nyland et al., 2004), the Strutter crutch, aimed to reduce neurovascular damage that is often a consequence of long-term crutch use. The rectangular design (Fig. 5) consists of two long rods in the axial direction and two short rods in the transverse direction serving as a base foot and underarm pad. Results indicated a significant difference in peak palmar forces on flat surfaces compared to an axillary crutch. User feedback indicated perceived stability was also increased, especially during the stance phase. Follow-up research reported an individual with dysfunctional myelodysplasia using the Strutter crutch indicated improved ambulation and physical health and the patient went from non-functional ambulation to functional community ambulation (Magee and Kenney, 2008). However, lower extremity parameters including walking speed, step length, joint angles, cadence, and energy cost did not show any significant difference.

Some designs place the impaired side on a saddle instead of transferring weight against the underarm or hands (TTilsley and Tilsleyilsley and Tilsley, 1998; Hunter, 2016). One example (Hunter, 2016), shown in Fig. 5, is a hands-free crutch design. This design's most significant parts are a saddle holding the injured leg in a bent position, a shaft, and a foot to replace the lower leg. However, experimental evaluations are needed to analyze how these designs function.

5. Concluding discussion

Crutch gait structurally alters gait features such as range of motion, ground contact point, velocity, distribution of pressures on the upper and lower limbs, etc. Percentage of partial weight bearing, velocity, and efficiency vary between crutch gaits. Therefore, each crutch gait and type of crutch should be prescribed based on individual needs and the physical capabilities of each patient. That is why we have summarized the important aspects of experimental research that studied crutches as major topics in the past four decades in Table 2.

Axillary and non-axillary are the two main categories of crutch designs. Axillary designs have not changed much over the past century, but need further modifications to reduce energy expenditure, increase comfort, reduce the risk of injury, and adapt to the impairment of individuals. Non-axillary crutches contain more variations in design. The forearm crutch is the most common non-axillary crutch with supporting angled-cuffs around the forearm instead of axilla pads under the arm.

Experimental studies have indicated that disabled crutch users and able-body subjects have different walking features and dynamics, thus healthy subjects do not completely represent the way a person with a disability would use the crutches. Disabled crutch users often have a lower range of motion, lower flexibility and hip movement, and greater physical strain. However, healthy subjects can be used during initial evaluations as long as the end-users are included throughout the design and experimental process.

A comprehensive crutch study should analyze and compare parameters from all categories to be able to draw conclusions. Crutch components that should be analyzed include the human parameters, crutch kinematics, and the interaction between them. Forces applied to the upper limbs and GRF are needed to understand the kinetics of crutch walking. Either oxygen consumption or heart rate can represent energy cost and efficiency. Gait variables include a vast range of parameters such as joint angles, step time, step length, velocity, and acceleration.

The evolution of crutches has frequently focused on modifying one part of the structure to improve one or more of these crutch parameters. Design modification of crutches has aimed to increase comfort, efficiency of walking, or stability by altering crutch handle, shaft, or tip. Peak forces applied to axilla and shoulder can be avoided by optimizing the crutch design and the prescribed gait. Energy efficiency can also be increased by adding shock absorbers to the shaft. However, there is a tradeoff between stability and efficiency due to change of height in springs. A symmetric rolling crutch tip has not shown a significant change in reducing pressure; however, a varying radius tip can alter the dynamic through redirecting forces.

Crutch research has evolved extensively over the past century. However, there has not been a unified method for studying crutch gait. This review has combined all the components of crutch studies and modifications of crutch design together. This paper aimed to show the importance of looking at multiple evaluation methods to standardize future studies. A comprehensive crutch study should compare and analyze components from all categories including forces, energy consumption, and gait variables.

Crutch design modifications need to focus on improving efficiency as well as comfort. Crutch gait should be considered as an inherently different gait that alters walking dynamics. Crutch users' self-selected speeds are commonly less that the most efficient crutch speed. Lack of training and design flaws are two contributing factors (Hall and Clarke, 1991; Mcbeath et al., 1974; Souza et al., 2010). Design modifications should increase performance while maintaining stability. Rehabilitation training should also prescribe crutch gaits based on the user impairment and upperbody strength.

Declaration of Competing Interest

Fatemeh Rasouli has no affiliations with or involvement in any organization or entity with any financial or non-financial interest in the subject matter or materials discussed in this manuscript.

Kyle Reed has two patents related to a device described in this manuscript. Specifically related to the Kinetic Crutch Tip (KCT), which is currently being sold as the MTip (https://meomtip.com). Both patents have licensing agreements with Moterum Technologies. The University of South Florida also has a financial interest in Moterum Technologies. A management plan has been implemented and followed to reduce any effects of these conflicts of interest.

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