CAST BRACE IN FRACTURES

ESSAY

SUBMITTED IN PARTIAL FULFILMENT

OF MASTER DEGREE OF

ORTHOPAEDIC SURGERY

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(1989)

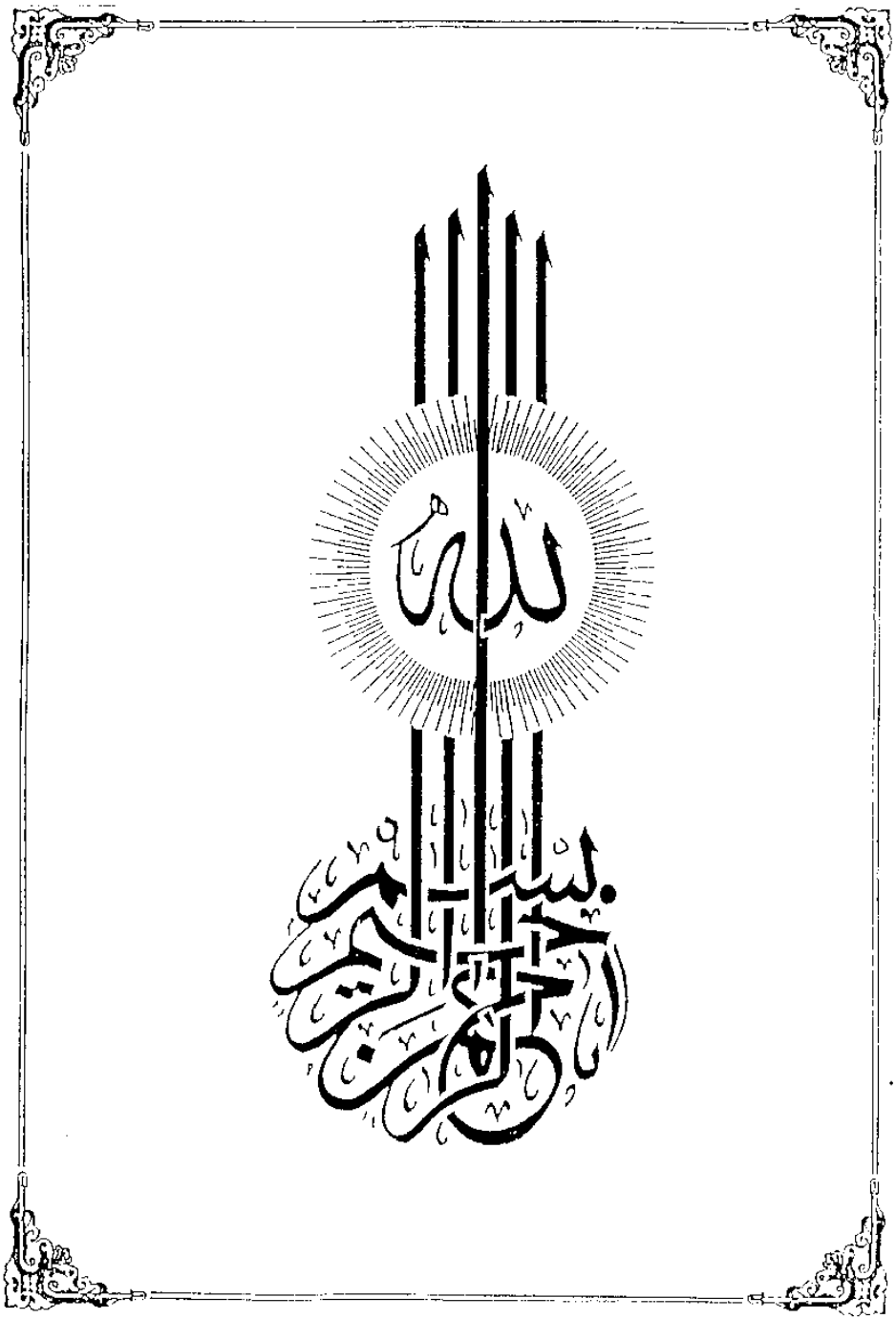
Acknowledgment

First and foremost, thanks are due to ALLAH the beneficent and merciful, who gives and facilitates every thing.

I wish to express my sincere appreciation and my deepest gratitude to *Professor Dr. Mohamed Nabil Khalifa*, professor of orthopaedic surgery, Ain-Shams University, for his constant support and careful guidance throughout the course of this work.

I am also grateful to *Professor Dr. Ibrahim Saied Darwish*, ass. Professor of orthopaedic surgery, Ain-Shams university, for his continuous support, valuable foresight and great encouragement.

Finally, I would like to thank all my colleagues, in the orthopaedic department, El-Sahel Teaching Hospital, for their unforgettable help and unfailing encouragement.



C O N T E N T S

HISTORICAL BACKGROUND

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The concept of fracture bracing is not new. Fracture bracing is a non-operative treatment of fractures using braces. This treatment is thought to have originated in the medical school of ancient China, and has been reintroduced by *Dehne, Sarmiento, Latta and Mooney*

It was in 1791 when *John Hunter* was confronted with a man with non union of a fracture in the proximal part of the femur instructed him to walk upon crutches and to press as much upon the broken thigh as the state of the parts would admit with a view to rouse the parts to action forcing them by a species of necessity to strengthen the limb. The result was rapid union of the limb allowing the man to walk with a cane within two months.

The first brace specifically designed for fracture care was described by *Smith in 1855* (Fig. 1.1). This device which had a waist band, an ischial support and a thigh lacer, was to be used for ununited fractures of the femur. *Smith* called the device a prosthesis. He believed that the device would substitute for the fractured limb. He also believed that containment of the thigh musculature in the laced cuff would help to maintain fracture alignment and much to his surprise, all seven fractures so treated were healed.

Fig. 1.1: Brace used by H.H.Smith.
(After Brown and Preston, 1985).

According to *Roper 1981*, the London Hospital Gazette of 1904 described the Hessing splint for the treatment of femoral fractures, this was a laced thigh and shin cuff made of leather joined by two lateral knee hinges.

In 1916 Delbert of France reported success in treating femoral fractures with an ischial weight bearing orthotic device.

Dehne initiated the weight bearing cast treatment for tibial fractures around 1950 in military personnel. Later reports in 1966 by him and his associates and *Witschi and Omer* had demonstrated that early ambulation of tibial shaft fractures in appropriately applied plaster cast is a method which almost uniformly results in union without significant complications. *Sarmiento and Sinclair* at 1967 used these principles in treating tibial shaft fractures in civilians.

Mooney and associates extended these principles further by reviving the hinged cast technique for the convalescent treatment of femoral fractures around 1970 (*Connolly et al., 1973; Brown and Preston, 1975; Roper, 1981*).

BIOMECHANICS OF CAST BRACE

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Initial experiences with patellar-tendon-bearing prostheses used by the below-the-knee amputees first led to the development of a below-the-knee cast which, moulded like a PTB prosthesis, would transfer weight-bearing pressures from the foot to the proximal tibia and patellar tendon virtually by passing the fracture site and therefore suspending the injured extremity. Results obtained with this functional cast were most encouraging as shortening was controlled and fracture healing took place uneventfully. Similar experiences were obtained with femoral and forearm fractures using plaster-of-paris casts moulded like quardrilateral ischial weight-bearing prostheses and Munster prostheses as used by the above-knee and below-elbow amputees respectively.

It soon became evident that none of these functional casts though effective in the management of fractures, were adequately transferring weight-bearing pressures to the proximal bony areas. Instead, the soft tissues of the extremities seemed to be responsible for the distribution of weight-bearing pressures and the maintenance of stability and alignment of the fractured bones (*Sarmiento, 1974*).

Electrical instrumentation of various braces were used to clarify this matter where braces were instrumented with pressur transducers located in their inner walls and placed in

areas opposite to the patellar tendon, the tibial flare, the crest of the tibia and the posterior aspect of the leg (Fig. 2.1). The data obtained consistently revealed a greater pressure reading over the gastro-soleus mass and significantly less over the bony prominences of the proximal tibia. It also revealed that the load bearing function of the brace was only fifteen percent which indicated that the limb carried over eighty percent of the load (*Sarmiento et al., 1974*).

This was supported by Meggitt et al. (*1981*) who showed that the thigh section of the femoral cast brace carries an average of ten to twenty percent of body weight during the healing of the fracture and does not function as a weight-relieving caliper but seems to have a bend-controlling function and thus allow safe intermittent compression of the fracture during walking. The three point fixation principle applies, with the upper thigh cast above and the shin cast below providing medial support, while the lower thigh cast and hinges give lateral support (Fig. 2.2). Antero posterior stresses are less because the muscles are balanced and there is a free fulcurum at the knee. The cast brace therefore; functions mainly as an "anti buckling hinged tube".

Wardlaw et al., (1981) claimed that in femoral fractures there was an increase in the fracture load as union progressed which was thought to be due to physiological feedback mechanism from the fracture site to the patient's central nervous system,

Fig. 2.1: Pressure transducers for measuring load distribution in fracture brace. (After Sarmiento et al., 1974).

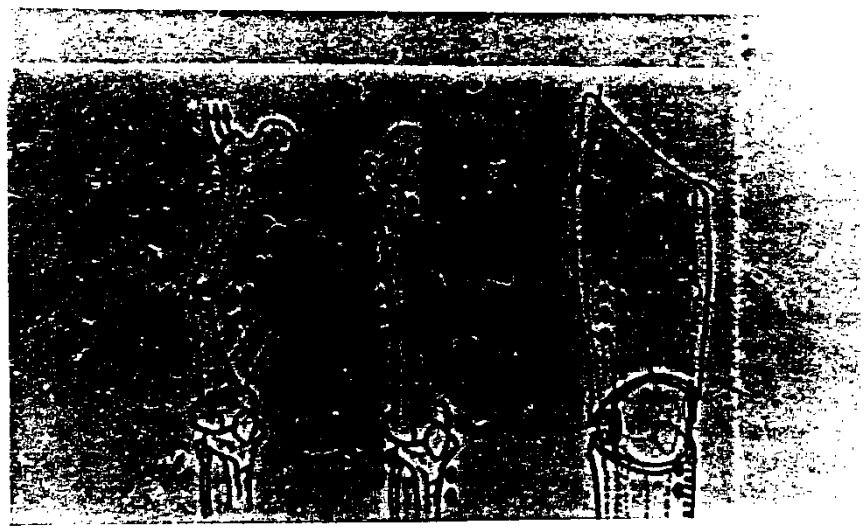

Fig. 2.2: The deforming forces in fracture of the shaft of the femur and the mechanism of three-point pressure in resisting these forces. (After Meggitt et al., 1981).

monitoring the stress the fracture will withstand.

Cineradiographic studies have indicated that the functional braces do not provide rigid immobilisation to the fractured bones as demonstrated by gross motion of the fragments under weight-bearing and functional conditions (*Sarmiento, 1974*). These movements, however, are fully elastic i.e. recoverable upon relaxation of load so that progressive deformity does not occur. The soft tissues of the injured extremity control the amount of motion occurring at the fracture site which is related to the fit of the brace and the extent of soft-tissue damage. In the case of the tibia and forearm, the interosseous membrane further seems to contribute to the stability of the injured parts.

The soft tissues have two major mechanisms for load bearing and provision of stiffness to the limb when encompassed in a fracture brace. The first mechanism is related to their incompressible-fluid like nature (*Sarmiento* and *Latta, 1981*). The muscle compartments act as a fluid-like structure surrounded by an elastic fascia container (Fig.2:3a). Dynamic load deforms the compartment of fixed volume (incompressible fluid) causing changes in their surface which stretch the fascial boundaries (Fig.2:3b). When these compartments are bound by a relatively rigid container such as a fracture brace, they can displace under load only until they have filled all the gaps within the container (Fig.2:3c). Once this slack is

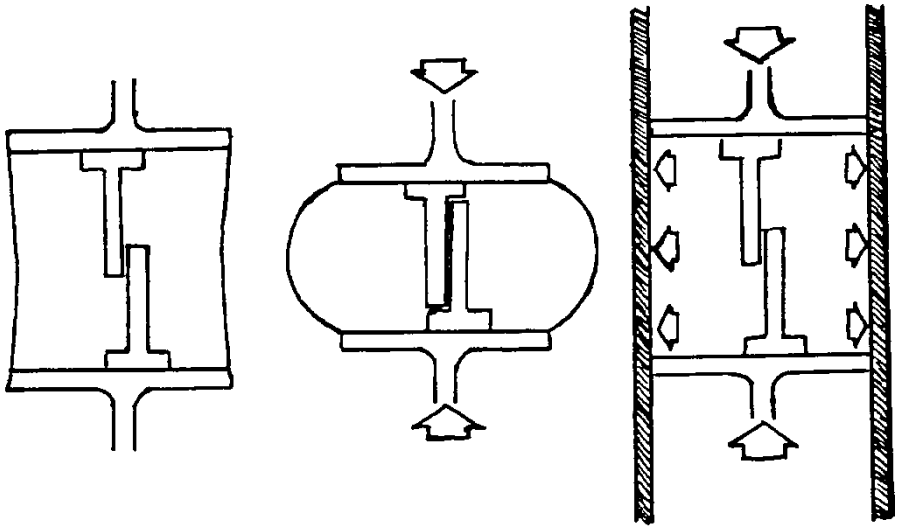

Fig. 2.3: A schematic depiction of the effect of a functional brace in stabilising a fracture by using the incompressible fluid-like nature of the soft tissues. (After Sarmiento and Latta, 1981).

taken up in the system, the muscle mass becomes rigid since its boundaries (the wall of the brace) do not move. After the load has been relaxed, the elastic fascial boundaries of each muscle return to their original shape which brings the fragments to their original position.

This mechanism of load bearing in the soft tissues which is called the "hydraulic effect of the tissues" is important in the early stages of management, when little healing has taken place in the bone or soft tissues. The fragments are loose and must rely heavily on the soft tissues for support until callus forms. The soft tissues must rely heavily on the degree of fit of the fracture brace in order for this mechanism to be effective. For long-term use, this mechanism cannot be relied upon since the dimensions of the soft tissues change with time through loss of oedema and atrophy . The fit of the brace cannot be maintained indefinitely. Loss of fit results in increased slack in the system which increases the displacement of the fragments required to produce mechanical equilibrium between the applied forces and the resistance of the tissues. This hydraulic effect of the tissues is not responsible for the long-term maintenance of length of the limb (*Sarmiento* and *Latta, 1981*).

The other soft tissue mechanism for load transfer involves their intrinsic strength in tension as they support the bone fragments at their natural attachments (*Sarmiento* and *Latta,*